BIOMECHANICAL EVALUATION OF POSTURAL STANCE STABILITY

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ABSTRACT

Postural balance, which refers to the essential ability of maintaining daily functions and sports activities, is one of the major concerns in society. As the static stance control serves as the indicator of dynamic control via assessment of postural sway, it is important to identify changes among different targeted groups. Therefore, the aims of this dissertation were to assess the changes of postural stance balance among young adults, middle-aged adults and transtibial amputees. Additionally, the aim was to examine the changes of muscular response to platform stiffness. The changes of postural control among young adults according to their body mass index classification and gender during bipedic stance and unipedic stance were investigated. The findings showed that obesity may lead to a poor postural control among young adults ($p < 0.001$). There is a tendency for females to have better postural performance than male. Then, postural control in bipedic stance and stance with toe-extension for young adults was examined. The findings demonstrated that the center of pressure displacement of stance posture with voluntary toe-extension does not differ with bipedic stance and no gender effect towards stance control was found. In addition, the changes of postural control and electromyography response on the support surface perturbation among healthy middle-aged adult and transtibial amputee were investigated. There is also no gender effect with human balance control found for both middle-aged adult and transtibial amputee. Both groups revealed a lower relative stability for direction control as the support surface perturbation inclined. There is left-right asymmetry of lower limb muscle response and displacement of center of pressure. Furthermore, the relationship of the anthropometric variable with postural control among transtibial amputee and age-matched able-bodied group were evaluated. The relative stability of transtibial amputee was negatively correlated with increased body mass index in static stance. No relationship was found in relative stability-body mass index and relative stability-body fat percentage in dynamic stance. A strong correlation between body weight two-point discrimination sensitivity test on trans-metatarsal (r = –0.787) and mid-foot (r = –0.784). The body mass index has a strong correlation with two-point discrimination sensitivity test on trans-metatarsal (r = –0.752) and mid-foot (r = –0.826). None of the variables such as age, weight, height, Q-angle, body mass index, skinfold measurement and foot sensations were found to be a predictor for postural control among TTA. The findings have eliminated the related variables as possible balance-related factors for transtibial amputees for future research. The findings could contribute to the understanding of the possibility of anthropometric variable impact on postural stability and development new rehabilitation program for transtibial amputee.

ABSTRAK

Keseimbangan badan merujuk kepada keupayaan penting dalam mengekalkan aktiviti harian. Ia merupakan salah satu kebimbangan utama di dalam kalangan masyarakat. Sebagai penuju kepada kawalan dinamik, kawalan statik adalah penting untuk mengenal pasti perubahan kestabilan badan di dalam golongan yang berbeza. Oleh itu, tujuan utama dalam tesis ini adalah untuk menilai perubahan keseimbangan badan di dalam kalangan orang dewasa muda, orang pertengahan umur dan pesakit transtibial. Selain itu, tujuan kajian ini adalah untuk mengkaji perubahan tindak balas otok-otok kepada pertukaran permukaan sokongan. Perubahan kestabilan badan di dalam kalangan orang dewasa muda dengan perubahan kategori indeks jisim badan dan jantina dalam keadaan pendirian ‘bipedic’ dan ‘unipedic’ telah dinilai. Ia menunjukkan bahawa obesiti boleh menyebabkan mengurangkan keupayaan pengawalan keseimbangan badan di kalangan orang dewasa muda (p < 0.001). Ia juga menunjukkan bahawa kecenderungan wanita dengan mempunyai prestasi keseimbangan badan yang lebih baik daripada lelaki. Kajian ini mengkaji tentang perubahan keseimbangan badan dalam keadaan pendirian ‘bipedic’ dan berdiri dengan ‘toe-extension’ di antara orang dewasa muda. Hasil kajian ini menunjukkan bahawa peranan ‘center of pressure’ dalam pengawalan keseimbangan badan dalam keadaan pendirian ‘toe-extension’ tiada perbezaan dengan postur pendirian yang ‘bipedal’, dan tiada pengaruhan untuk faktor jantina dalam kawalan keseimbangan badan. Selain itu, kajian ini telah menyelidik perubahan keseimbangan badan dan tindak-balas aktiviti elektromiografi dalam situasi sokongan permukaan pendirian dinamik di antara orang dewasa yang pertengahan umur dan pesakit transtibial. Tiada pengaruhan dengan faktor jantina dalam kestabilan badan. Ia juga mendedahkan bahawa permukaan yang lebih curam akan mengurangkan ‘relative’ kestabilan seseorang. Terdapat asimetri aktiviti otok-otok kaki untuk kaki kiri dan kanan. Tambahan pula, hubungan unsur–unsur antropometri dengan kawalan postur antara pesakit transtibial dan orang dewasa yang pertengahan umur telah dinilai. ‘Relative’ kestabilan pesakit transtibial adalah diperkaitkan dengan perubahan indeks jisim badan dalam pendirian statik. Tiada bukti menunjukkan bahawa hubungan dalam ‘relative’ kestabilan–indeks jisim badan and ‘relative’ kestabilan–peratusan lemak badan adalah berkaitan dalam pendirian dinamik. Kajian ini mendapati bahawa berat badan adalah berkaitan dengan ‘two-point discrimination test’ di ‘mid-foot’ (r = –0.787) dan tumit kaki (r = –0.784). Selain itu, indeks jisim badan adalah berkaitan dengan ‘two-point discrimination test’ di ‘mid-foot’ (r = –0.752) dan tumit kaki (r = –0.826). Tiada pemboleh ubah seperti umur, berat badan, ketinggian, Q- sudut, indeks jisim badan, ukuran ‘skinfold’ dan sensasi kaki telah ditemui sebagai peramal untuk pengawalan keseimbangan badan. Kajian ini telah menghapuskan pemboleh ubah yang berkaitan dengan keseimbangan badan dalam golongan pesakit transtibial untuk penyelidikan masa depan. Hasil kajian boleh menyumbang lebih pemahaman kepada kemungkinan pengaruhan antropometri terhadap keupayaan kestabilan sistem badan dan informasi untuk program yang baru dalam pemulihan bagi pesakit transtibial.

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LIST OF SYMBOLS AND ABBREVIATIONS

%	:	Percentage
º	:	Degree
N	:	Newton
cm	:	centimeter
Hz	:	Hertz
kg	:	kilograms
min	:	minute
mm	:	millimeter
s	:	seconds
A leg	:	Amputated leg
AMPPRO	:	Amputee mobility predictor with the use of prosthetic
Amp	:	Amputee
ANOVA	:	Analysis of variance
AP	:	Anterior-posterior
APSI	:	Anterior-posterior stability index
BBS	:	Biodex balance system
BK	:	Below-knee
BLS	:	Bipedic stance
BMI	:	Body mass index
BoS	:	Base of support
CNS	:	Central Nervous SFystem
CoG	:	Center of gravity
CoM	:	Center of mass
CoP	:	Center of pressure

DCL	:	Directional control
EC	:	Eyes closed
EMG	:	Electromyography
EO	:	Eyes open
FES	:	Falls Efficacy Scale
FFA	:	First-fitted amputee
LEA	:	Lower-extremity amputee
LoS	:	Limits of Stability
MCT	:	Motor control test
ML	:	Medial-lateral
MLSI	:	Medial-lateral stability index
MS	:	Medium stump
MTP	:	Metatarso-phalangeal
NA	:	Non-amputated
OSI	:	Overall stability index
PVD	:	Peripheral vascular disease
RMS	:	Root mean square
SOT	:	Sensory organization test
SPU	:	Skilled user
SWT	:	Standing without toe-extension
SS	:	Short stump
TFA	:	Transfemoral amputee
TTA	:	Transtibial amputee
ULS	:	Unipedic stance
VN	:	Van ness amputee

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CHAPTER 1: INTRODUCTION

1.1 General

Postural balance is a critical skill for daily activity. Postural balance improves circulation and reduces stress, physically and mentally. An upright erect or bipedal posture is the natural and basic locomotion pattern for human. A twelve-month old toddler starts learning and applying body balance skills by crawling (Shumway-Cook & Woollacott, 2007). After that, he learns how to stand still by controlling his trunk via his subconscious. After toddlers are able to stand still independently, they automatically try to move by walking, jumping, and running. These continuous steps of action represent the nature of human balance posture.

Poor balance may lead to critical issues such as falling risks to general health. As the second largest public health problem, falling is defined as the action of inadvertently coming to rest on the ground or another lower level, excluding intentional change in position to rest on furniture, a wall, or other objects (World Health Organization, 2008). Falls are prominent among external causes of unintentional injuries as a result of poor balance control in response to perturbation. According to the report of the National Health Interview Survey, it is estimated that the United States has 13,042,000 people who experienced a fall in the year 2010 (Adams, Martinez, Vickerie, & Kirzinger, 2011). Fall represent a major risk to the elderly. Approximately 28-35% people aged 65 years and over have experienced a fall each year (World Health Organization, 2008). For people aged 75 years and above, 30% of adults are in fair of poor health, and 35 – 42% of them may suffer moderate or severe accidental injuries, such as bone fracture or mortality (Resnick & Junlapeeya, 2004). 87% of bone fracture cases in adults occur through standing height falling (Court-Brown, Garg, & McQueen, 2001). During year 2010, the total cases of non-fatal fall injuries reported in United States were estimated at 43 people per 1000. Different group of falling rate were estimated: aged under 12 years (0.042 %), aged 12 to 17 years (0.061 %), aged 18 to 44 years (0.026 %), aged 45 to 64 years (0.043 %), aged 65 to 74 years (0.055 %), and aged 75 and above (0.115 %) (Adams et al., 2011). Thereby, it is statistically proven that risk of fall injury in elderly is about 2 to 3 times higher than the young population. The fracture risk was associated with age, body weight, height and BMI in the normal elderly population (Compston et al., 2014). About 30 – 50% of elderly fall cases are caused by unintentionally slipping and tripping. In addition to the general population, loss of balance control would also impact on disease groups (for example: individuals with stroke and osteoporosis). Finally, post fall syndrome may be another side effect of falling. The symptoms of post fall syndrome are dependence, loss of autonomy, confusion, immobilization, depression, loss of self-confidence, and limited balance control which might change the quality of life (World Health Organization, 2008).

1.2 Postural Stability among Transtibial Lower-Extremity Amputees

Lower extremity amputation is defined as surgical procedure that refers to removal of part or all of the lower limb. Transtibial amputation, where part of the below knee (BK) part of the leg removed, is the most common performed major limb amputation. There are several indications for amputation, including trauma, peripheral vascular disease (PVD), infection, tumors, neurological disorders, and congenital deformity (Magee, 2008). The younger population tends to experience more malignancy, congenital anomalies, and trauma-related amputation, whereas 75% of the older population experience the vascular disease-related amputation (Fitzpatrick, 1999). Figure 1.1 showed the levels of transtibial amputation. The amputation site may occur through joints or bone (Shurr & Cook, 1990). For below knee amputation, the amputation surgery can be performed at various levels of lower leg include very short BK, short BK, medium BK, and long BK.

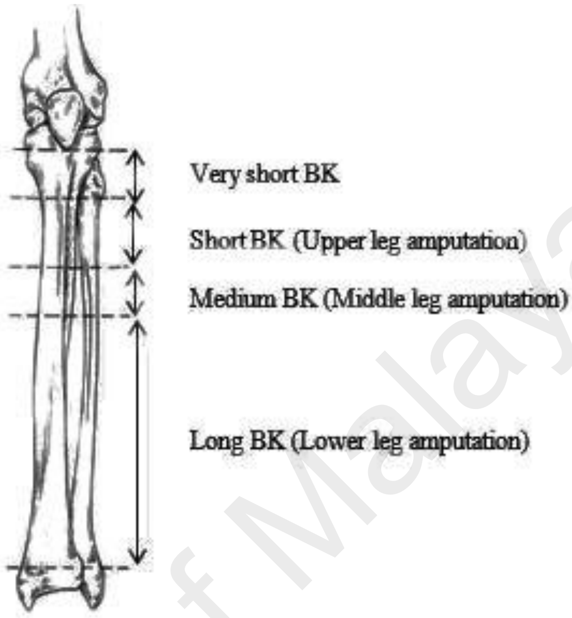

Figure 1.1: Common levels for partial lower limb amputation. (Adapted from Magee (2008))

Although the balance control mechanisms in lower-extremity amputees (LEA) are different from the general population, the contributing elements to postural control are similar. For transtibial amputees (TTA), the two knee joints work as a strong mechanism to regulate human postural transitions. However, the loss of anatomical ankle tends to change the locomotion mechanism from the bipedic stance pattern (Vrieling et al., 2008). Studies showed that LEAs have limited mobility, greater metabolic demands, and high incidences of pain and discomfort compared with the general population (Klute et al., 2009). Due to the loss of ankle joint in the lower extremity, the range of foot placement is affected and the balance recovery strategy does not provide sufficient force to maintain postural stability at the amputated side, so a new adaptation mechanism is developed to compensate for the ankle joint. Other challenges such as adaptation to prosthesis and balance confidence should be overcame by the LEA to improve their quality of life.

1.3 Research Problem Statement

The postural balance ability is a critical concern since it is a basic ability for every human being. People learned to balance the body in a natural way since they are born. There are balance-related factors that could affect balance performance during daily life. This research provides significant and new information for postural balance in erect posture, further altering the gait mechanism. There are some groups of individuals who have difficulty balancing their body such as patient with cerebral palsy, osteoporosis or amputees. Although many studies related to the body's postural balance have been done, there are still some unknown factors that could be related to postural instability that have not been found. In order to find out those balance related factors for healthy and unilateral transtibial amputee, the understanding of postural stability is essential. The findings of the study presented in this research could potentially be useful in the field of rehabilitation. Rehabilitation in balance may provide opportunities for regaining balance by correcting the body posture with the right techniques.

1.4 Research Objectives and Aims

This study aims to identify anthropometric variables that affect postural stability. In order to realize this aim, several objectives were identified. The objectives of this research study are outlined as follows:

i. To assess the postural activity among healthy young adults according to their BMI classification and gender during bipedic stance and unipedic stance.

ii. To determine the changes of postural stability in bipedic stance and stance with toe-extension standing posture.
iii. To investigate the muscular response on the postural activity during the support surface perturbation during middle-aged adulthood.
iv. To identify the changes to postural control and lower limb muscle activation in TTA during support surface perturbation.
v. To examine the possible balance-related factors associated with postural balance of TTA during stance posture.

1.5 Research Approach

This research is an attempt to identify balance-related factors in healthy individual and TTA. Balance-related factors are typically approached with experimental testing where the effect of physiological factors were investigated in the research. The changing of the balance mechanism in static postural balance presented that the postural balance of amputees is more unfavorable than in healthy individuals. It is essential to note that the outcomes of balance-related factors with postural balance can be used in future rehabilitation for physicians, prosthetists, and physical therapists. The present research provides new findings that can be used as comparative or improved knowledge for rehabilitation technique. Moreover, this research provides baseline data for the comparison of upright stance patterns and muscle activity in middle-aged adults. The baseline of stance control could be used for the rehabilitation assessment of the independent functions in dynamic control. Based on the objective outline above, the research has considered relevant literature, tools, techniques and methods to conduct clinical experiments on the healthy population and transtibial amputees. The assessment test has provided a better understanding of body performance and activity level. The research was mainly focused on laboratory-based human testing. A relevant literature review provides the rationale of the study and help to gather sufficient information and knowledge on the postural stability of the healthy population and transtibial amputees.

1.6 Thesis Organization

This thesis consists of 8 chapters that are briefly as below:

Chapter 1 provides an overview of the study field, research objectives, scope, and the approach of the postural balance. The importance of the study is discussed.

Chapter 2 briefly reviews the literature associated with an emphasis on the experimental studies of human standing in healthy individuals and transtibial lower-limb amputees. It summarized the important findings of the previous studies for healthy individuals and transtibial amputees.

Chapter 3 presents the effect of body mass index impact on postural control among young adults during bipedic and unpedic stance.

Chapter 4 contains the comparison of postural control in bipedic stance and stance with toe-extension conditions.

Chapter 5 examines the postural stability and muscle activation under the circumstance of support surface perturbation among middle-aged adults. The impact of platform stiffness on the directional control and muscle activity for left-right leg is also encountered.

Chapter 6 investigates the changes of muscle response and postural performance with support surface perturbation among transtibial amputees. The experiments involve different levels of support surface perturbation for the limits of stability test.

Chapter 7 identifies the balance-related factors that influence the postural performance of transtibial amputees.

Chapter 8 summarizes the major findings from the research works. Conclusions are drawn and future works that are related with postural stance control are recommended in this section.

CHAPTER 2: LITERATURE REVIEW

2.1 General

In order to establish a better understanding for this study, this chapter provide a comprehensive reviews of previous literature on the general population and transtibial amputees. General knowledge concerning standing balance stability is introduced. The detail of possible balance-related factor is listed and explained. In addition, the previous findings of static balance for healthy adult and transtibial amputee are reviewed in this chapter.

2.2 Background of Static Equilibrium

Research in human biomechanics plays a significant role in advancing the knowledge and ideas of human mobility for public health, sport science, rehabilitation, and clinical practice. The term "postural balance" refers to the fundamental element to maintain daily life activities. Postural balance is the ability to regulate the body through orientating the center of mass (CoM) within the base of support (BoS) and achieve the body equilibrium against perturbation (Alexandrov, Frolov, Horak, Carlson-Kuhta, & Park, 2005). In the literature, postural balance is usually measured to study the basis of movement. In order to study human balance, kinematic data must be collected for precise measurement. The static posture is defined as the alignment of body segments in stance conditions (Figure 2.1). CoM is a virtual point that is equivalent to the total body mass at which the average of mass distribution for each segment is assumed to be concentrated, against disturbance (Hernández, Slider, Heiderscheit, & Thelen, 2009). BoS is the displacement area of the center of pressure (CoP) of the net ground reaction force against the ground surface which is confined to act within it (Barrett, Cronin, Lichtwark, Mills, & Carty, 2012). The standard deviation of the instantaneous position where the resultant of all ground reactions forces act under the feet and reflects the neuromuscular response to correct the CoM displacement is defined as CoP (Collins & Luca, 1993; Lenka & Tiberwala, 2010). The line of gravity for the ideal posture alignment will pass through earlobe, cervical vertebrae, the tips of shoulder, thorax, lumbar vertebrae, posterior to the hip joint, anterior to the knee joint, and anterior to the lateral malleolus (Kisner & Colby, 2002). The trajectory of CoP is a widely used parameter to quantify human balance.

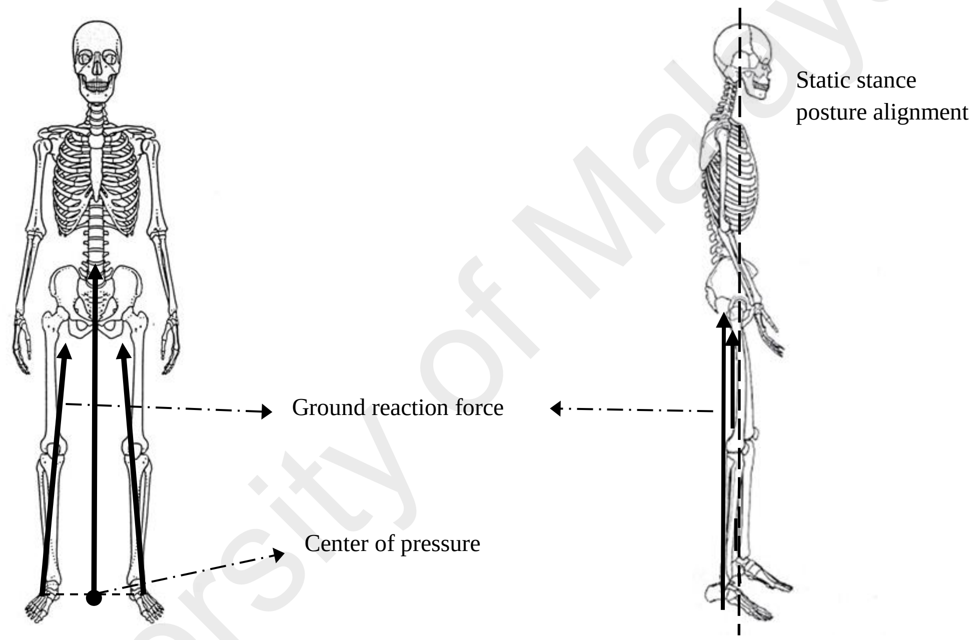

Figure 2.1: The ideal posture alignment for static equilibrium. The distribution of forces during the control of quiet stance. (Adapted from Kendall, McCreary, Provance, Rodgers, and Romani (2005))

Static balance control is served as a balance predictor for dynamic control via postural sway (Ku, Abu Osman, & Wan Abas, 2014). Postural sway is defined as the deviation in CoP displacement on the supporting surface. A greater postural sway represents the body trunk required more effort to achieve body equilibrium. The development of new measurement techniques and technologies have improved the understanding of postural sway in body stability for all populations.

2.2.1 Newton's Laws of Motion

The fundamentals of human biomechanics allow us to calculate unknown forces. When a person moves and stops, forces are exerted, either internally and externally, on the human body. The theory of force was first published by Isaac Newton with three laws included: law of inertia, law of acceleration and law of action-reaction. These three laws have formulated the basic concepts and classical mechanical characteristics of forces acting on objects (McLester & Pierre, 2007). With the theories of laws of motion, both the magnitude and direction of the forces exertion can be calculated.

Newton's first law

Newton’s first law also known as the law of inertia, stated, “*Every object in a state of uniform motion tends to remain in that state of motion unless an external force is applied to it*”. This law states that a particle will remain in rest state if it is at rest, while the moving object will remain with constant velocity if there are no external forces applied on it (Glenn Research Center, 2015). When sum of all force from vertical and horizontal directions is equal to zero, the object will remain balance with a constant velocity. If there are any external forces applied, they will change the velocity of the object. The gravitational force is an external force that is applied on all objects (Hall, 2007).

Newton's second law

Newton’s second law, also known as the law of acceleration, stated “*The direction of the force vector is the same as the direction of the acceleration vector”*. It developed a relationship between an object’s mass (m), its acceleration (a) and applied force (F) (Richards, 2008). This law indicated that a particle acted upon by an external force moves such that the force is equal to the time rate of change of the linear momentum (mass x acceleration). According to Equation 2.1 for Newton’s second law, the increased mass will generate greater force in order to maintain the same acceleration. Likewise, with a constant object mass, the change in force will change velocity, while the change in velocity will generates a force. As every static and dynamic movement has a force, Newton’s second law can be used for the force calculation in human biomechanics.

$$F = ma \tag{2.1}$$

Newton's third law

Newton’s third law also known as law of action-reaction, stated “*For every action there is an equal and opposite reaction*”. This law indicated that each force exerted by one object on another is counteracted by a comparable force exerted by the second object on the first object (Glenn Research Center, 2015). In this law, the gravitational force should be encountered for all actions and reactions. When an individual is standing still on a level surface, forces will be applied downward to the ground and another opposite force will be exerted from the ground to equalize the force magnitude. In this case, the opposite forces are known as ground reaction forces (Richards, 2008).

2.3 The Nature of Postural Control Mechanism

There is a complex system to sustain the body position in a stable condition. The postural control is a complex system that consists of a sensory part, processing part and an effectuating part which all could be affected by either sleepiness, fatigue accumulating throughout the day or hormone levels (Jorgensen et al., 2012). A disorder in any of the condition, or any combination of these subcomponents may leads to postural instability. Figure 2.2 shows the physiological and neurophysiological factors that are related to the alteration of postural balance.

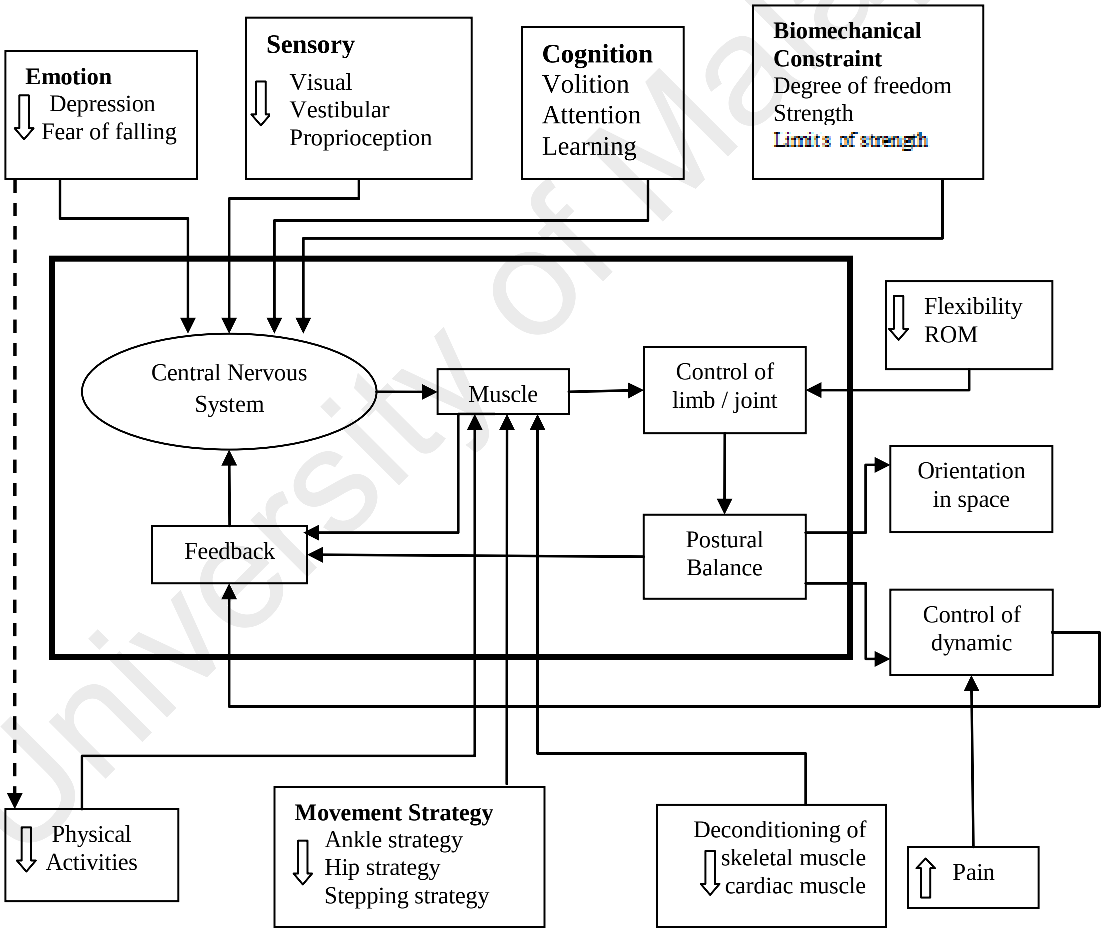

Figure 2.2: The factors that may associated with postural balance. This diagram is modified from the postural-related factor of Horak (2006) and Hausdorff et al. (2001). The shaded box illustrates the interaction of locomotor system and certain age-related factors in physiological capacity that may mediate postural stability.

The central nervous system (CNS) is used for active control, while stabilization of the body is considered to be controlled by the passive mechanical structures (Masani, Vette, Abe, Nakazawa, & Propovic, 2011). Horak (2006) suggested that six sub-components are required to retain postural balance, including biomechanical constraints, movement strategies, sensory strategies, orientation in space, control of dynamics, and cognitive processing. Studies of sensory strategies for balance control demonstrated the vital role of integrations between the visual, vestibular, and proprioception elements in quiet standing. The acquisition of motor control through the central nervous system stimulates the reflective output of movement, where the incoming information obtained from the sensory systems is transmitted to the nervous system. Balance instability may possibly be induced by pathophysiologic and aging processes, behavioral, pharmacologic, or environmental factor (Abolhassani et al., 2006; Lord, Sherrington, & Menz, 2001).

2.3.1 Central Nervous System

The CNS is part of the processing system that mainly coordinates activity, integrating information, and regulating mechanisms in the human body. The CNS is composed of the brain and spinal cord as shown in Figure 2.3. The spinal cord functions as the pathway that receives and transmits nerve impulses from the receptors to CNS. It also helps in reflexive postural control, and processes the proprioception information from the joints, skin and muscles, and the reflexes via motor neurons (Drake, Vogl, & Mitchell, 2009; Netter, 2010). The nerve impulses relay from the receptor or sense organ towards the CNS through afferent neuron, while the impulses that leave the CNS are transmitted toward the effectors through efferent nerves. The brain is the center of human nervous system. The brain is composed of multiple specialized areas that integrate information. The brainstem is located at the region which connects between the spinal cord and cerebrum. It consists of the midbrain, medulla oblongata and the pons. The brainstem receives proprioception inputs from the skin and head muscles, and regulates life supporting automatic activity such as breathing and sleep. The cerebellum is located behind the brain stem and is responsible for coordination and balance. It helps in modulating the movement responses from obstacles. The sensory modalities, visual, vestibular, and proprioception systems, have been known as the key physiological components of balance control (Karimi, Ebrahimi, Kahrizi, & Torkaman, 2008).

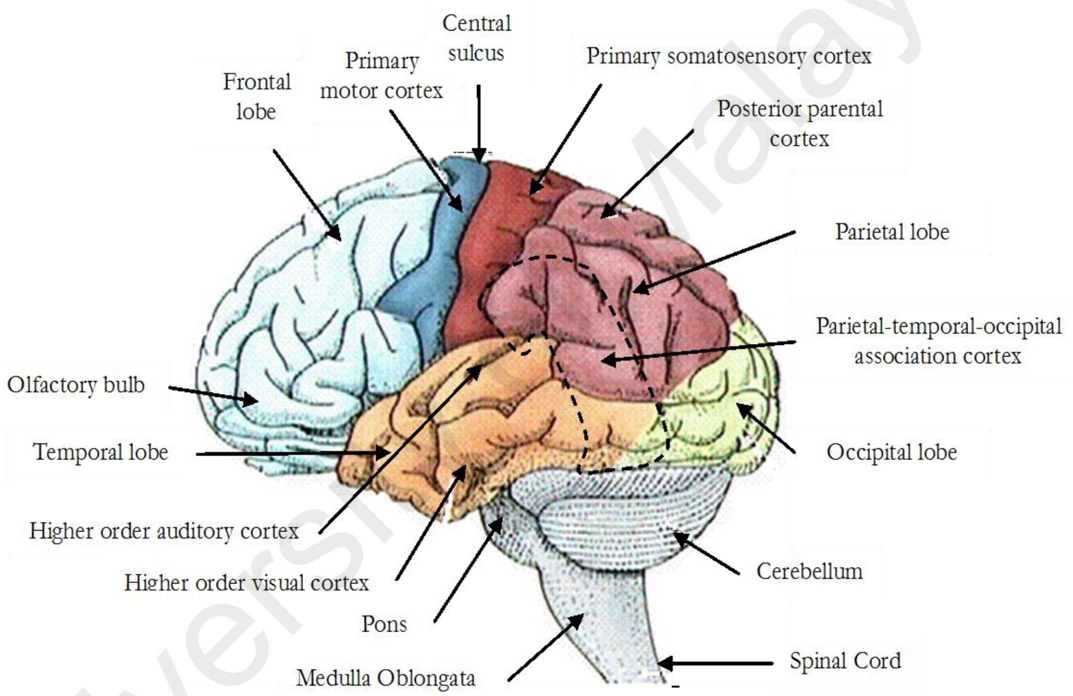

Figure 2.3: The anatomy of central nervous system. (Adapted from Drake, Vogl, and Mitchell (2009))

2.3.2 Sensory Strategy

2.3.2.1 Visual System

The vision system plays a role in fall prevention. The vision system detects obstacles in the surrounding environment and transmits relevant information to help the body in balance preservation (Drake, Vogl, & Mitchell, 2009). Vision also allows virtual distance estimation and physical view for balance recovery in body adjustment against perturbation. The eye is an organ that is designed to focus the image from the surrounding environment on the retina with high precision (Netter, 2010; Shumway-Cook & Woollacott, 2007). Light enters the eye through the cornea, the transparent front part of the eyes. The cornea focuses and bends light then passing through the pupil, the hole-sizing controlled by the iris, in order to control the amount of light enter the eye. The light then travels through the transparent gel-like substance called 'vitreous humor' and is focused on the retina. The retina contains two types of photoreceptor: the rods and the cones (Netter, 2010). The rods are functional for vision at night as they are it is only sensitive to shorter wavelengths compared to cones that are more sensitive to normal daylight vision. The fovea is the blind spot of the eye which has no photoreceptor, therefore it cannot detect any light and form images. Then the bipolar cells and ganglion cells that connected with the optic nerve fiber relay vision information to the CNS (Netter, 2010; Shier, Butler, & Lewis, 2008).

Therefore, visual inputs are generally used in controlling low frequency disturbances and dynamic tasks (Gill et al., 2001; Tomomitsu, Alonso, Morimoto, Bobbio, & Greve, 2013). The eyed closed condition will directly increase CoP displacement, CoP velocity and velocity moment (Yoon, Yoon, Shin, & Na, 2012). Hafstrom, Fransson, Karlberg, Ledin, and Magnusson (2002) stated that stance balance with eyes open in darkness does not differ from having eyes closed. They speculated that expected visual feedback information in the eyes open condition may delay the postural strategy in a dark environment. The studies of Grangeon, Gauthier, Duclos, Lemay, and Gagnon (2015) also revealed that it is more challenging to control sitting stability than standing stability under the unsupported eyes closed condition. Figure 2.4 illustrated the testing conditions of the sensory organization test, which is designed to examine visual, proprioceptive and vestibular feedback.

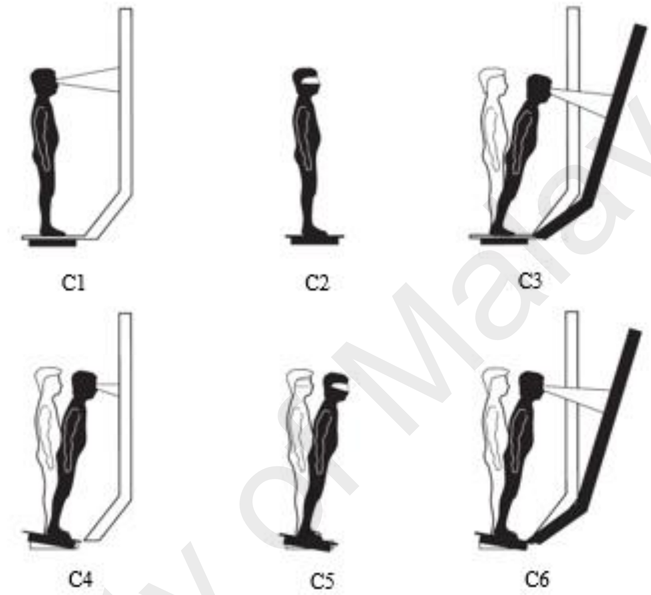

Figure 2.4: Sensory organization test with six testing conditions (C1, eyes open surround fixed with platform fixed; C2, eyes blindfolded with fixed platform; C3, eyes open surround sway-referenced with platform fixed; C4, eyes open with surround fixed and platform sway-referenced; C5, eyes blindfolded with platform sway-referenced; C6, eyes open with surround sway-referenced and platform sway-referenced. (Adapted from Steindl, Kunz, Schrott-Fischer, and Scholtz (2006))

2.3.2.2 Vestibular System

The vestibular system, it is a component located in the inner ear that control balance performance and sense of movement in daily activities. This system is sensitive to the position of the head in space and sudden changes in the direction of movement of the head (Netter, 2010; Shier, Butler, & Lewis, 2008; Shumway-Cook & Woollacott, 2007). The system is also crucial for the coordination of motor responses, and vestibular inputs may help to stabilize the eyes. The vestibular system can be divided into peripheral and central components (Netter, 2010). The peripheral component consists of the sensory receptor and the eighth cranial nerve. The vestibular system is part of the membranous labyrinth of the inner ear which located in the temporal bone. There are five receptors included in the vestibular system: three in semicircular canals, utricle and saccule (Shier, Butler, & Lewis, 2008). The semicircular canals consist of three canals, horizontal, anterior, and posterior canals. These semicircular ducts functions as angular accelerometers to detect rotary head movement. The utricle and saccule are the otolith organs that serve to provide information for the body orientation and linear acceleration of the head; these components are sensitive with the gravity and acceleration (Shier, Butler, & Lewis, 2008). When the head lies horizontally, the macula of the utricle in horizontal is in a rest state. The eighth cranial nerve is referred as the vestibule-cochlear nerve or auditory vestibular nerve. It is a function for body equilibrium and sound transition from the inner ear to the CNS (Seeley, Stephens, & Tate, 2006).

The central part have four vestibular nuclei within the medulla oblongata which included: the lateral vestibular nucleus (Deiters'), the medial vestibular nucleus, the superior vestibular nucleus (Bekhterev's), and inferior vestibular nucleus (Drake, Vogl, & Mitchell, 2009). The lateral vestibular nucleus receives information from utricles, semicircular canals, cerebellum, and spinal cord and help in activate the antigravity muscle group in neck, trunk and limb (Shumway-Cook & Woollacott, 2007). The input information of medial and superior vestibular nucleus is from semicircular canals which helps in associating the motor nuclei in the eye muscle to stabilize gaze during head motion. The inferior vestibular nucleus receives inputs from utricles, saccule, semicircular canals, and the cerebellar vermis.

2.3.2.3 Proprioception

The proprioception system is an automatic sensitivity mechanism that sends impulses and signals through the CNS for the information on location and strength of effort that is applied in movement (Shier, Butler, & Lewis, 2008; Shumway-Cook & Woollacott, 2007). Sensation and perception is the aspect that involves different stages of information which are detected by a variety of sense organ from the surroundings such as sound, smell and image. Sensation is considered to be a physiological process which involves various sense organs, receptors, neural pathways, and regions of the brain (Drake, Vogl, & Mitchell, 2009). Perception is the psychological process by which people derive and interpret these sensation data (Drake, Vogl, & Mitchell, 2009). It is a platform that allows the sensory system to receive raw physical energy as information from the environment.

The role of sensation in movement control could be help in modulating the output of movement that generates the activity of pattern from the spinal cord (Seeley, Stephens, & Tate, 2006). The transduction process converts the information into sensory neural impulses. All the sensory neural impulses in movement control are transmitted through the ascending pathways, which contribute to movement patterns. The ascending pathway is defined as the path that transmits information from the trunk and limbs to the sensory cortex and cerebellum (Netter, 2010). There are two systems that divide the ascending pathway: the dorsal column-medial lemniscal system and the anterolateral systems. The dorsal column-medial lemniscal system will send information from muscles, tendons and joint sensibilities for touch and pressure receptors to the higher brain center (Shumway-Cook & Woollacott, 2007). The anterolateral system which consists of spinothalamic, spinorecticular, and spinomesencehalic tracts that serve as the fiber to transmit impulses, are responsible for sending information on pain, temperature, crude, touch and pressure receptors to the brain center (Drake, Vogl, & Mitchell, 2009; Netter, 2010). The somatosensory cortex is a processing area for all the proprioception components. The somatosensory cortex can be categorized as the primary somatosensory cortex and secondary somatosensory cortex. The primary somatosensory is the area where information from joint receptors, muscle spindles, and cutaneous receptors are integrated, thereby the information about movement in a given body area exists. Contrast sensitivity is an aspect for movement control, since it allows the detection of the shape and edges of the objects (Seeley, Stephens, & Tate, 2006). There are interactions between perception and feedback. In addition to regulate the dynamic restraints, the motor control, proprioception system also indirectly influences dynamic joint stability and muscle stiffness (Riemann & Lephart, 2002). The receptors for proprioception that are located in muscle, tendons, joints and feet are used to determine joint position, muscle force and orientation of limb (Patestas & Gartner, 2009). The cutaneous receptors aid in reflex movement and provide information for the orientation of body position. The receptors help postural standing by detecting information regarding the orientation of the ankle and foot pressure distribution (McComas, 1996). The ankle proprioception provides information for the ankle position adjustment and upper body movement for complex motor tasks (Han, Anson, Waddington, Adams, & Liu, 2015).

Horak (2006) reported that healthy persons who stay in well lit-environment with a firm BoS will rely on 10% of vision, 20% vestibular and 70% proprioceptive information in postural control. If they stand on an unstable surface, the sensory weighting will incline to the aspect of vision and vestibular systems while reducing the dependence on the proprioceptive inputs of postural orientation (Peterka, 2002). People with peripheral vestibular loss or proprioception loss from neuropathy are limited in their ability for postural sensory dependence and thus, suffer risks of falling that are related to the sensory response (Horak, 2006; Jáuregui-Renaud, 2013). The feedback of sensory systems would also be altered in association with aging (Figure 2.5).

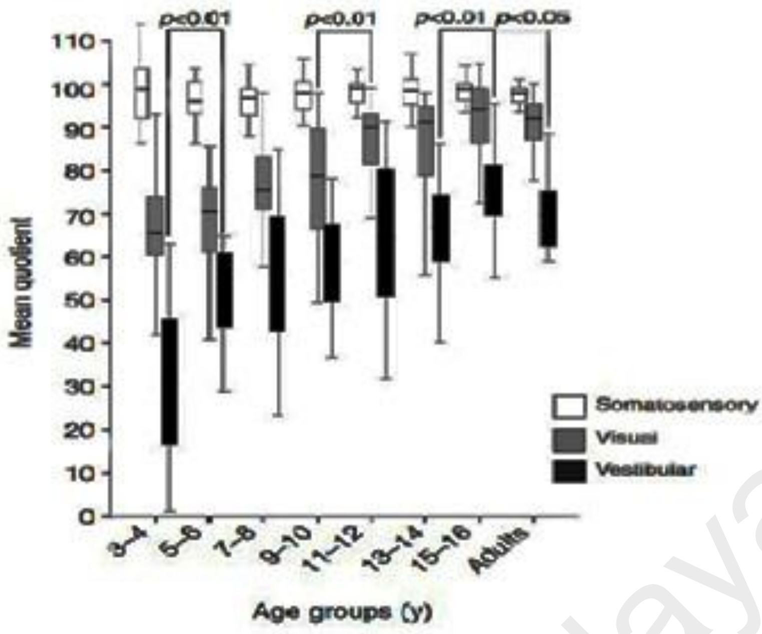

Figure 2.5: The changes of postural control on visual, vestibular and proprioception for different age groups. (Adapted from Steindl et al. (2006))

2.3.3 Biomechanical Constrains

For biomechanical constraints, also known as whole-body mechanical consideration, the human postural control system accommodates a variety of factors to preserve the equilibrium against external disturbances such as degree of freedom, strength, and limits of stability (Horak, 2006). Among all the factors, the most important factor involves the size and quality of the BoS (Tinetti, Speechley, & Ginter, 1988). The BoS is defined as the area around the body's supporting parts (hands or feet) on the supporting surface (McGinnis, 2005). A wider area of BoS will improve body stabilization. Figure 2.6 illustrates that there are three basic types of BoS in human movement. The area that is shaded with a dashed line represents the area of BoS.

Individuals who are prone to fall tend to have a small limits of stability and area of BoS. The ML instability is strongly related with fall risks for the elderly population (Maki, Holliday, & Topper, 1994). The postural system will automatically adapt in accordance with task requirements to ensure the CoM relocates within the BoS of the feet. When the vertical line of gravity moves outside the BoS, a torque that evoke the angular motion of the body is generated and disruptes body stability.

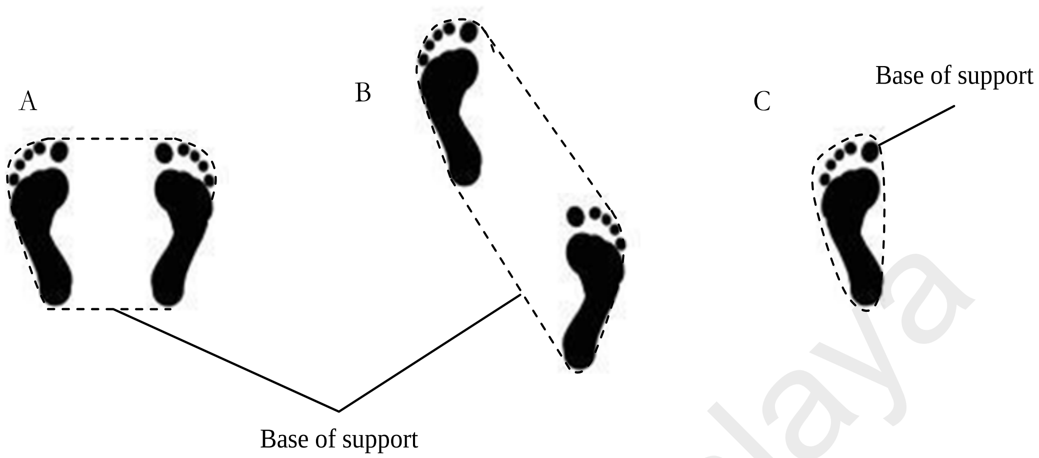

Figure 2.6: Area of base of support. (A) double-leg (B) double-leg with step leg (C) Single-leg. (Reproduced from Gschwind et al. (2014))

Sport activities that require a high level of postural balance depend on the BoS area and the location of CoM (Ackland, Elliott, & Bloomfield, 2009). When the area of BoS becomes severely smaller, the allowable motion for CoM in the transverse plane and ML stability will be decreased (Chapman, 2008; Mehdikhani, Khalaj, Chung, & Mazlan, 2014). On the contrary, a widening BoS also will limit ankle mobility, natural frequency and increase ML biomechanical stiffness (Kirby, Price, & MacLeod, 1987; Winter, 1995). The height of CoM that is relative to the area of BoS may also affect the ability of body to balance, as the position of CoM gets nearer to the ground, the area of BoS increases for balance recovery. If the CoM does not stay within the BoS, the body will experience a continuous state of imbalance in dynamic balance (Shumway-Cook & Woollacott, 2007). In spite of the fact that foot placement will not affect the measurements of static foot posture, a standardized foot placement is recommended to ensure a high level of measurement consistency (McPoil et al., 2014). Yoon et al. (2012) suggested that the posture of standing still where2 the width of foot placement is equal to half of the shoulder width for a high reliability balance capacity.

2.3.4 Movement Strategies

The movement of the body segment is vital for the body balance. The mass distribution of each body segment influence the body equilibrium at the daily activities (Del Porto, Pechak, Smith, & Reed-Jones, 2012; Hernández et al., 2009). For the activities such as sitting, standing, or walking, the different postures used for these activities may require different orientations of body segment to prevent falls. The human body has postural movement strategies that include ankle, hip and step strategies for postural control by regulating the CoM in the sagittal plane for balance perturbation (Winter, 1995) (Figure 2.7). It involve the muscle synergies, contact forces, and movement patterns joint torques (Horak, Henry, & Shumway-Cook, 1997). The ability of generating sufficient joint torques helps to achieve balance stability of the trunk by actively shifted the weight bearing between the two extremities.

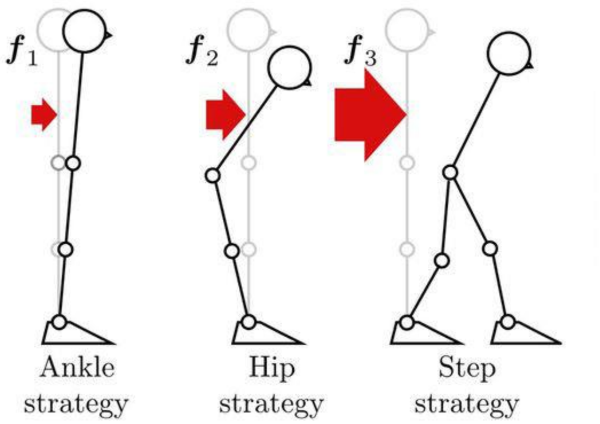

Figure 2.7: Effect of recovery strategies on different force of perturbation. The recovery strategies are used by normal adults for regulating upright sway. The level of forces applied is $f_1 < f_2 < f_3$. (Reproduced from Kisner and Colby (2002))

In human anatomy, all muscle groups mentioned are vital in the postural control system, as the gastrocnemius muscle refers to the muscle group that located in the back part of the lower limb. The hamstring muscles refer to 3 different types of posterior thigh muscles: semitendinosus, semimembranosus and biceps femoris, while the paraspinal muscle group is located next to and roughly parallel with the human spine. It consists of a small muscle that is attached to the vertebrae. It is responsible for control of the range of motion of the bones, as well as assisting with the motions of the whole trunk. With help from paraspinal muscle group, it limits spine angular motion, which avoids overextension injuries to the spinal cord. All the skeletal muscles normally work contrary to one another, as one muscle is in the contraction state while the opposite muscle is in the relaxed state.

As a small perturbation, when the vertical line of gravity falls slightly in front of the knee joint, ankle strategies are applied. In the ankle joint strategy, the activation of muscle in ankle occurs from distal to proximal and induces the torques in the ankle joint (Winter, 1995) (Figure 2.8). This strategy mainly depends on the information from the proprioception system. The activation of gastrocnemius muscles generates plantar flexion torque against the angular motion in the forward direction (Kuo & Zajac, 1993). At the same time, the paraspinal muscle and hamstring muscles are activated simultaneously to regulate the forward motion of the hip and knee joint, further affecting the proximal body segment. The illopsoas plays a role in preventing the occurrence of hypertensions at the hips. For the activation of the thoracic erector muscle group, it helps in limit the forward motion and shifts the trunk in the backward direction (Horak & Nashner, 1986; Nashner, 1976; Shumway-Cook & Woollacott, 2007). In addition, when the BoS is reduced, it affects the performance of the ankle strategy in regaining balance (Horak & Nashner, 1986).

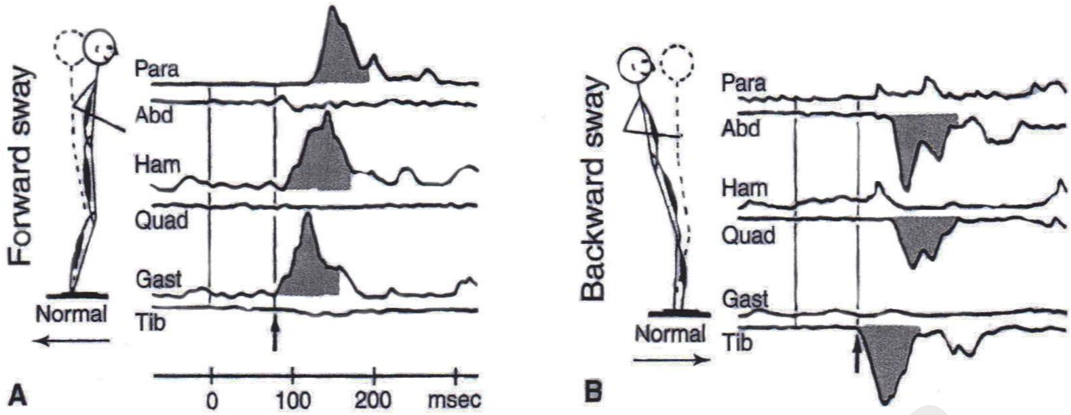

Figure 2.8: The muscle synergy activation for ankle strategy against perturbation in (A) forward displacement (B) backward displacement. (Adapted from Horak and Nashner (1986))

If a larger disruption of balance occurs in the forward direction, the hip strategy is applied (Figure 2.9). In the hip strategy, activation of the hip and trunk muscles occurs and induces the torques at the ankle joint, knee joint and hip joint (Winter, 1995). This balance recovery strategy mainly depends on the information from the vestibular system. The abdominals and quadriceps femoris muscles are activated and as a result the trunk shifts in a backward direction in order to restore the displacement of CoM against the larger perturbation (Horak & Nashner, 1986; Salsabili, Bahrpeyma, Forogh, & Rajabali, 2011). A well-developed in musculoskeletal system may also help in body adjustment against perturbation. Deterioration in the aspect of locomotion, muscle control, and tone strength may influence postural control (de Oliveira, de Medeiros, Frota, Greters, & Conforto, 2008).

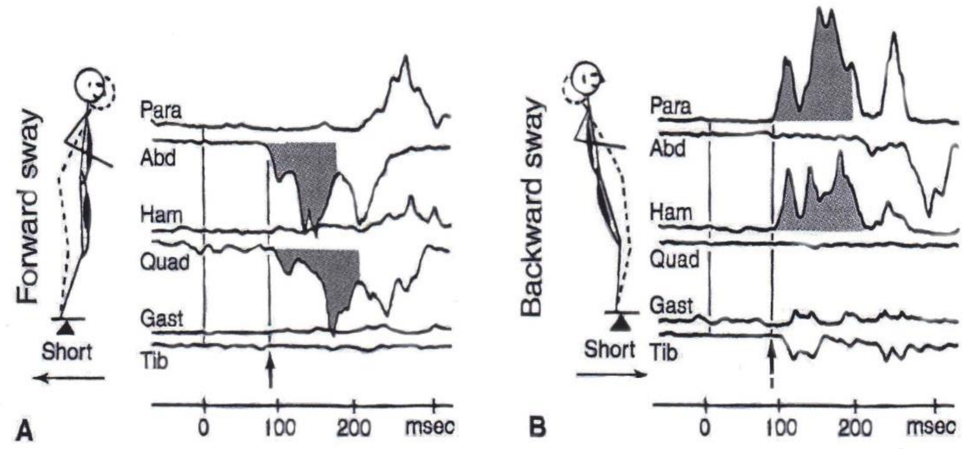

Figure 2.9: The muscle synergy activation for hip strategy against perturbation in (A) forward displacement (B) backward displacement. (Adapted from Horak and Nashner (1986))

There are studies that describe that the ankle joint is vital for body sway in AP direction and the ankle response is to keep the whole body in a vertical position during a quiet stance. Whereas, hip placement is responsible for keeping body equilibrium by regulating more in the ML direction and hip response is excellent for faster and larger CoM perturbation (Gatev, Thomas, Kepple, & Hallett, 1999; Salsabili et al., 2011). The active hip abductors which are controlling pelvic obliquity and ML trunk control are crucial for postural stability (Andrysek et al., 2012). Apparently, small external perturbation may depend on the compensatory of ankle response for ankle feedback control. The inclination of perturbation tends to induce hip feedback control and reduce the ankle feedback response (Park, Horak, & Kuo, 2004). Generally, the ankle and hip strategy are used more as the basis of mobility, while the step strategy is used for rapid transition in the dimension of BoS with respect to CoM changes.

2.3.5 Orientation in Space

The ability of body orientation in space relies on perception, gravity force, contact surface and visual input. With the input information from the visual, vestibular and proprioception systems, the CNS will automatically compare, process, and integrate the information. The CNS will determine the body position and alter body segments to orient the alignment of CoM for the related body parts return to an equilibrium state. The body is oriented according to the angle of the ground contact surface, as the trunk is perpendicularly over the ground surface for a bipedic stance. Healthy people are able to tolerate the influence of gravitational vertical in the dark within 0.5º (Horak, 2006). Indeed, the ability of body adjustment to the vertical gravitational in the dark is depend on the perception in vertical postural (Bisdorff, Wolsley, Anastasopoulos, Bronstein, & Gresty, 1996). Thus, the internal representation of visual system inclines an angle in persons with unilateral vestibular loss whereas the internal representation of postural leads to the decline of the ability in postural alignment, result to the body imbalance.

2.3.6 Control of Dynamics

Postural stability can be affected by the control of dynamic balance during walking and posture switching. A quiet stance is a complex task that involves the integration of multiple body segments, joints and sensory systems (Ku, Abu Osman, Yusof, & Wan Abas, 2012b). Compared to quiet stance, the balance control over a gait cycle and posture switching require a more complex system. The weight is shifted from one leg to the other and CoM will alter and shift out of the area of BoS during functional walking. Studies stated that dynamic postural stability in the forward direction is caused by the position of the swing limb under the falling CoM, while the stability in lateral direction is caused by the lateral trunk control (Horak, 2006; Prince, Corriveau, Herbert, & Winter, 1997).

Figure 2.10 illustrates the changes of CoM position during different stages of human walking. The elderly that are prone to falls tend to have a larger lateral excursion of body CoM and more irregular lateral foot placements (Teasdale & Simoneau, 2001). It seems that the balance control of the elderly will narrow the step lengths and step width to negotiate obstacles, and will depend on central reorganization to increase balance control (Woollacott & Tang, 1997).

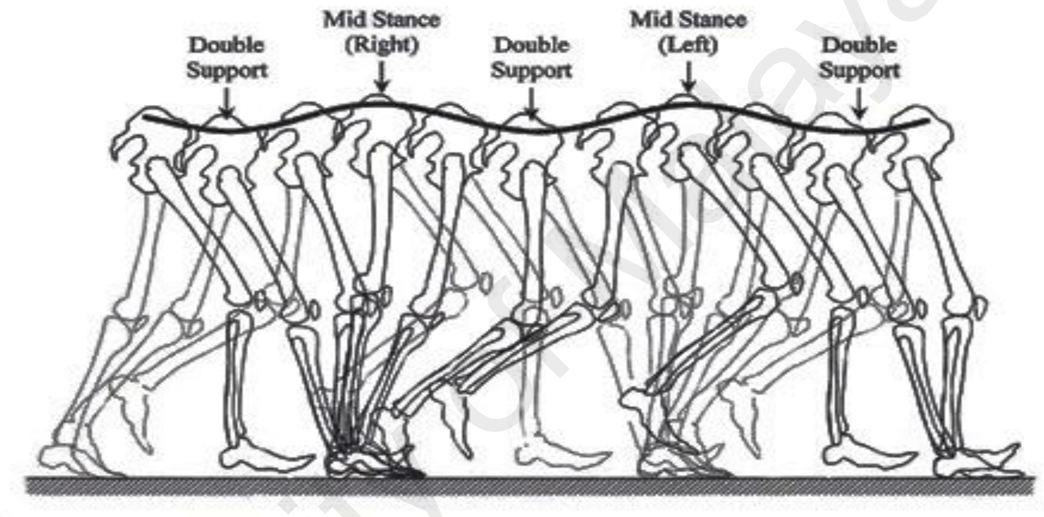

Figure 2.10: The vertical displacement of CoM during ambulation. The changes highest elevation and lowest elevation of body CoM occur at the stage of mid stance and double support respectively. (Adapted from Rose and Gamble (2006))

2.3.7 Cognitive Processing

The cognitive processing is required in the postural control mechanism. Cognitive processing in balance involved the attention and ability of learning of a person. Mignardot, Olivier, Promayon, and Nougier (2010) stated that obese people require more attentional resources to regulate postural stability than non-obese people during unipedal stance (Mignardot et al., 2010). As with the learning ability, it is a process which resulted from experiences and practices. The effect of learning is a relatively permanent change. Higher levels of neural processes in cognitive are the basis for adaptive and anticipatory aspects in postural control (Shumway-Cook & Woollacott, 2007). Cognitive information may increase the reaction time in response to a task (Teasdale & Simoneau, 2001). A more complex postural task may require more in term of cognitive processing. Thus, the reaction time and performance in cognitive tasks rises as the difficulty of a postural task increases (Horak, 2006).

2.3.8 Other factor

2.3.8.1 Age

Age is indirectly related to balance stability. The postural stability of children will improve with age. When people are getting older, their balance posture and gait will change accordingly. Previous studies consistently demonstrated that the aging process will impair balance stability (Abrahamova & Hlavačka, 2008; Medina, 1996). The aging process is associated with changes in adipose tissue, bone mass, muscle mass, muscle strength, power, and range of motion. Decline in muscle mass and bone mass diminishes functional mobility and balance, and increases the falling risk. Medina (1996) stated that the reduction of muscle mass in lower extremities with age is greater than in upper extremities. Forty percent of muscle strength in lower extremities can be reduced from ages between 30 to 80 (Aniansson, Hedberg, & Henning, 1986). Older individuals tend to have a higher risk of hip fracture than young population (Butler, Norton, Lee-Joe, Cheng, & Campbell, 1996). With no musculoskeletal impairment, the impact of aging on postural control could be minimized (Alexander, Shepard, Gu, & Schultz, 1992). In addition to the musculoskeletal system, the aging process also deteriorates the sensory systems (visual, vestibular, and proprioception systems) for body motion, and delays the postural response and reaction time against perturbation (Abrahamova & Hlavačka, 2008;

McIlroy & Maki, 1996; Yoon et al., 2012). As the loss of spinal flexibility is related to the changes of postural alignment, the decrease of range of motion and loss of spinal flexibility may lead to a characteristic flexed posture. Therefore, the elderly are more likely to rely on visual feedback for balance control, while children are more sensitive to proprioceptive and pressoreceptor information (Hytonen, Pyykko, Aalto, & Starck, 1993).

2.3.8.2 Body Mass Index and Somatotype

Body Mass Index (BMI) is a commonly-used index of relative weight (Keys, Fidanza, Karvonen, Kimura, & Taylor, 1972). It is defined as body weight (kg) per square meter of body height. According to the categories set by the World Health Organization, it is classified into four classifications: underweight (< 18.50 kg/m^2), normal weight (18.50 – 24.99 kg/m^2), overweight (25.00 – 29.99 kg/m^2), and obese group (> 30 kg/m^2). Obesity is generally acknowledged as an excessive of body mass, fat accumulation or adipose tissue distribution that could increase health risks and decrease occupational and recreational comfort as a result of injuries such as sprains, strains and dislocations (Matter, Sinclair, Hostetler, & Xiang, 2007). Previous studies found that obese individual have a more significance decrease in postural balance and suffer higher risks of falling than other BMI groups (Greve, Alonso, Bordini, & Camanho, 2007; Lord & Sturnieks, 2005; McGraw, McClenaghan, Williams, Dickerson, & Ward, 2000). The increase of body weight will impose a new biomechanical constraint (Blaszczyk, Cieslinska-Swider, Plewa, Zahorska-Markiewicz, & Markiewicz, 2009). Hue et al. (2007) reported that the functional abilities were reduced with increased body mass. They also suggested that balance stability may improve with decreasing body mass. Dutil et al. (2013) found the postural balance of obese individuals has higher postural sway compared to normal weight group under the eyes closed (EC) condition. Besides the comorbid disorders, obesity may occur together with mental health disorders such as depression which further may indirectly affect the confidence level regarding mobility (Klinitzke, Steinig, Blüher, & Wagner, 2012).

Sheldon’s somatotype theory is used to distinguish the human body shape which can be classified into 3 basic body types: ectomorph (slim and linear type), mesomorph (muscular type), and endomorph (round or fat type) (Miller, 2014) (Figure 2.11). An individual can have a combined type of body shape. There are several regions where adipose tissue normally distributed. The adipose tissue distribution for the android type (apple-like body shape) is concentrated in the thorax-abdominal region, while for the gynoid type (pear-like body shape) adipose tissue is usually found around the hip and thigh areas (Clark, 2004). Generally, the android type will occur in males and gynoid type occur in females. Obese females tend to accumulate a greater amount of adipose tissue in the lower extremities. Adipose tissue distribution can increase the difficulty of regulating CoM and CoP at the BoS, further affecting postural stability. Moreover, the increase of body height will exert larger CoP displacement; while increase of body weight is related to the ankle torque and noise generated in feedback control (Era et al., 2006).

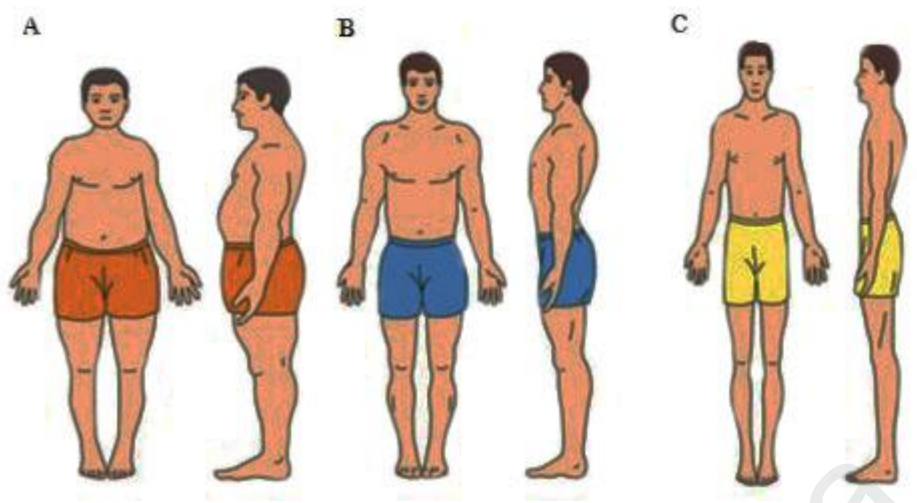

Figure 2.11: The basic body type of somatotype. (A) Endomorph (B) Mesomorph (C) Ectomorph. (Adapted from Gledhill et al. (2010))

2.3.8.3 Circadian Rhythms

An individual uses almost 12-hour per day for the daily activities. The magnitude of human biological rhythms is related with the changes to the environment. Environmental factors such as daylight, body temperature, social interactions, and times of meals could influence the performance of postural balance. Jorgensen et al. (2012) demonstrated that circadian rhythm affected the postural balance of the elderly population. The postural control in the morning is better than in the afternoon or evening (Gribble, Tucker, & White, 2007). Besides that, sleepiness is related to the circadian rhythm where a sleepy individual will generate higher postural sway. Goel, Basner, Rao, and Dinges (2013) revealed that the circadian rhythms have significant effect on cognitive performance, attentional demands and physiological alertness for the parameters of psychomotor vigilance test performance during an 88 hour period of limited to no sleep among healthy adults (Figure 2.12 – 2.13).

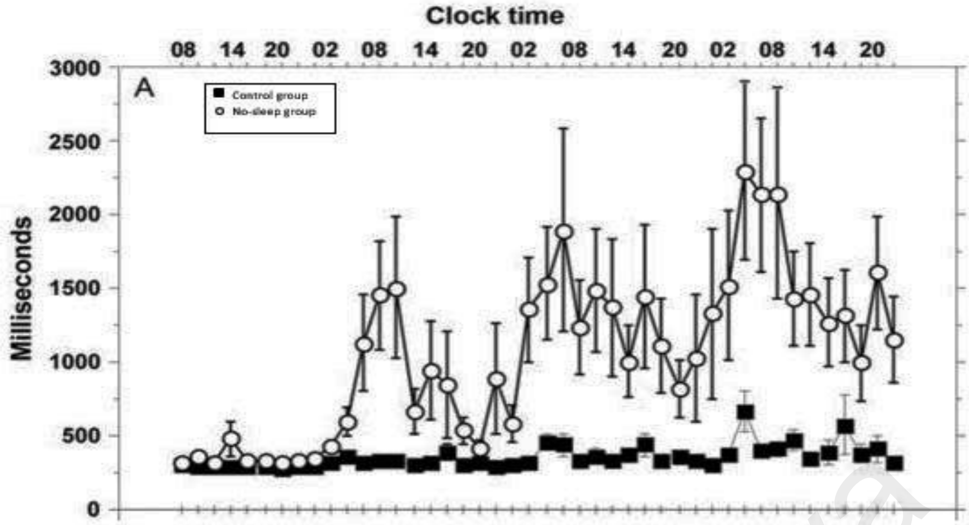

Figure 2.12: The mean of reaction times for psychomotor vigilance test that > 500ms was considered as frank errors of omission and referred to as lapses of attention (i.e., responding too slowly). (Adapted from Goel et al. (2013), with permission)

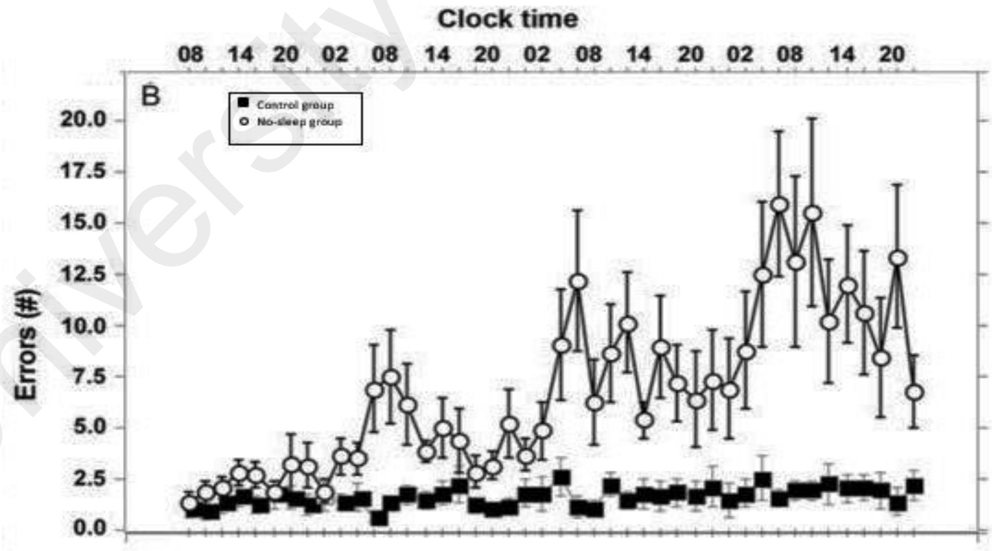

Figure 2.13: The mean of error of commission for psychomotor vigilance test that results from premature responses and reflect impulsiveness (i.e., responding too fast). (Adapted from Goel et al. (2013), with permission)

2.4 The Recent Findings Regarding to Balance Control in Lower Extremity Amputee

As the missing ankle results in changes to the balance mechanism, postural stability among TTA can be investigated through several methods. Horak (2006) reviewed the interaction and contribution among various physiological systems with balance. Although his findings are only limited for the general population, the basis of postural stability for mobility between normal individuals and TTA is still similar. The next section will discuss the findings regarding postural stability of lower-extremity amputee in the last 40 years. This content was reproduced from the study Ku, Abu Osman, and Wan Abas (2014), with the agreement of all authors.

2.4.1 Research Methods

There are various approaches implied for studies of balance among lower extremity amputee in quiet standing (Ku, Abu Osman, & Wan Abas, 2014). From the review of studies in recent years, there are studies that assessed the balance ability in static standing posture (Duclos et al., 2009; Duclos et al., 2007; Fernie & Holliday, 1978; Gaunaurd et al., 2011; Hlavackova et al., 2011; Lenka & Tiberwala, 2010; Mayer et al., 2011; Mouchnino et al., 1998; Nadollek, Brauer, & Isles, 2002; Rougier & Bergeau, 2009; Vittas, Larsen, & Jansen, 1986) and the ability of balance in dynamic standing posture (Andrysek et al., 2012; Aruin, Nicholas, & Latash, 1997; Barnett, Vanicek, & Polman, 2012; Bolger, Ting, & Sawers, 2014; Curtze et al., 2012; Kolarova et al., 2013; Molero-Sánchez et al., 2015; Nederhand et al., 2012; Viton et al., 2000; Vrieling et al., 2008). There are other studies that examined the ability of balance in static and dynamic standing posture (Buckley, O'Driscoll, & Bennett, 2002; Clark & Zernicke, 1981; Hermodsson et al., 1994; Quai, Brauer, & Nitz, 2005; Vanicek et al., 2009). The tested groups for the reviewed studies were divided into trauma (Bolger, Ting, & Sawers, 2014; Buckley, O'Driscoll, & Bennett, 2002; Clark & Zernicke, 1981; Curtze et al., 2012; Duclos et al., 2009; Duclos et al., 2007; Gaunaurd et al., 2011; Hermodsson et al., 1994; Hlavackova et al., 2011; Molero-Sánchez et al., 2015; Mouchnino et al., 1998; Nederhand et al., 2012; Rougier & Bergeau, 2009; Viton et al., 2000; Vittas, Larsen, & Jansen, 1986; Vrieling et al., 2008), vascular (Andrysek et al., 2012; Aruin, Nicholas, & Latash, 1997; Barnett, Vanicek, & Polman, 2012; Clark & Zernicke, 1981; Curtze et al., 2012; Gaunaurd et al., 2011; Kolarova et al., 2013; Nadollek, Brauer, & Isles, 2002; Nederhand et al., 2012; Quai, Brauer, & Nitz, 2005; Vittas, Larsen, & Jansen, 1986; Vrieling et al., 2008), cancer (Andrysek et al., 2012; Duclos et al., 2009; Duclos et al., 2007; Mayer et al., 2011; Vrieling et al., 2008) and limb deficiency (Curtze et al., 2012; Gaunaurd et al., 2011). The number of amputee participants ranged from 5 to 47. The average year since amputation for the participants ranged from 1.4 to 23.8 years. The detail of demographic characteristics of reviewed studies of quiet stance among LEAs is presented at Appendix B.6.

Table 2.1 gives an overview of different methodological protocols which related to quiet stance of LEAs over the past 20 years. Methodological challenges particularly increased with the interpretation of balance components as standard protocols were affected by many dependent variables: sample size, sample duration, foot distance, and difficulty of task performed. For example, the LEA studies reported that the distance of foot placement varied in a broad range. However, a broad stance was demonstrated to facilitate greater postural control in quiet standing (Bonnet, 2012). The placement distance between patients' feet ranged from 1 cm (Vittas, Larsen, & Jansen, 1986) to 30 cm (Aruin, Nicholas, & Latash, 1997). Discrepancies in foot distance may influence the reliability of the results. Besides, Gaunaurd et al. (2011) assessed the innominate inclination measurement in lower-extremity balance control. Most of the studies utilized two to three repetition trials in recording the balance performance.

As the upright standing posture is a complicated task for amputees, three main aspects are used to measure the human standing posture: (1) body segment displacement, (2) muscle activity, and (3) movement pattern of CoM and CoP (Balasubramaniam & Wing, 2002). There are different methods for assessing CoP, the force plate, which involves measurement of CoP displacement with transducers and assesses the pattern of balance variability during standing posture, is commonly used. In addition to using the single force plate, balance can also be evaluated by the integration of force plate with other systems (Aruin, Nicholas, & Latash, 1997; Buckley, O'Driscoll, & Bennett, 2002; Clark & Zernicke, 1981; Hermodsson et al., 1994; Mouchnino et al., 1998; Nadollek, Brauer, & Isles, 2002; Quai, Brauer, & Nitz, 2005; Viton et al., 2000; Vittas, Larsen, & Jansen, 1986), which including Vicon System (Andrysek et al., 2012; Curtze et al., 2012), Zebris FDM-S Forceplate System (Hlavackova et al., 2011), NeuroCoM System (Barnett, Vanicek, & Polman, 2012; Duclos et al., 2009; Duclos et al., 2007; Vanicek et al., 2009), and Computer-Assisted Rehabilitation Environment System (Vrieling et al., 2008).

In some studies, balance skills were evaluated using functional assessment, such as Berg Balance Scale (Duclos et al., 2009; Duclos et al., 2007), Sensory Organization Test (Barnett, Vanicek, & Polman, 2012), Community Balance and Mobility Scale (Andrysek et al., 2012), Activities-Specific Balance Confidence scale (Vrieling et al., 2008), Amputee Activity Score (Vrieling et al., 2008), Functional Reach Test (Quai, Brauer, & Nitz, 2005), and Visual Analogue Scale (Nadollek, Brauer, & Isles, 2002). The selection of postural control measurement tools will not affect the potential variables. In fact, only a few of the studies reported on the ICC value for the measured variables. The test-retest reliability was emphasized in order to validate the reliability of the statistics. High reliability indicates that the variable is strongly associated with postural sway, whereas low reliability implies a weaker effect on sway.

Table 2.1: Studies reporting balance control in lower-extremity amputees.

Study	Groups	Protocol	Instrumentation	Sampling rate (Hz)	Outcomes measures	Findings
Molero-Sánchez, Molina-Rueda, Alguacil-Diego, Cano-de la Cuerda, and Miangolarra-Page (2015)	TTA, control	Stand still with footwear and the hands placed alongside	SMART EquiTest System	100	MXE DCL MVL	TTA < control TTA < control (for backward) TTA < control (for certain direction)
Bolger, Ting, and Sawers (2014)		Stand still with standard shoe (3/8"heel height)	Vicon motion capture system	N	CoM sway CoP magnitude	TTA = control TTA = control
Kolarova, Janura, Svoboda, and Elfmark (2013)	TTA	Stand still with standardized foot position. Limits of stability test movement.	SMART EquiTest System	100	MXE DCL MVL	TTA < control TTA < control TTA = control
Andrysek et al. (2012)	TFA,VN, control	Intervention for 4 weeks with 20 min each session. Stand still with footwear and hand placed alongside. The feet placed align their heels. 3 trials performed. 1min for each trial.	Vicon MX^C system, Wii Fit	1200	CoP excursion CB&M score	TFA > VN > Control TFA < VN < Control
Barnett, Vanicek, and Polman (2012)	TTA	Intervention for 3 sessions. Stand still with flat footwear. Duration 8 s. Limits of stability test movement. Fall effect.	NeuroCom Equitest®	100	SOT	Changes occur in utilizing proprioception and visual inputs
					LOS	Increased in endpoint CoG excursion and directional control
Curtze et al. (2012)	TTA, control	Stand still with feet parallel. The feel set to 20% the height about the hip width. 3 trials performed. Fall effect.	Vicon system and force platform	1000	Ankle joint moment	TTA : NA leg > A leg Control : = between 2 legs
					Hip joint moment	TTA : NA leg = A leg Control : = between 2 legs
Nederhand et al. (2012)	TFA,TTA	Stand still with hands placed alongside. Feet placed 20 cm apart. 3 trials performed and 90 s for each trial. 1-2 min for resting. Dynamic movement.	Computer-controlled dynamic motion platform	360	Load distribution	NA leg > A leg

Table 2.1, continued: Studies reporting balance control in lower-extremity amputees.

	Groups	Protocol	Instrumentation	Sampling rate (Hz)	Outcomes measures	Study
Gaunaurd, Gailey, Hafner, Gomez-Marin, and Kirk-Sanchez (2011)	TFA, KD	Stand comfortably with base of support of 10 cm.	Laser-Line ™	N	Limb length PII Hip extension	NA leg > A leg NA leg < A leg NA leg > A leg
Hlavackova, Franco, Diot, and Vuillerme (2011)	TFA	Stand still with hands placed alongside. Feet separated 10 cm. 3 trials performed and 30 s for each trial. Resting for 1 min.	Zebris FDM-S Multifunction Force Measuring Plate system	100	CoP amplitude CoP velocity Regularity of CoP	NA leg > A leg NA leg > A leg NA leg < A leg
Mayer et al. (2011)	TTA	Stand still with feel parallel at hip width. 2 trials performed and 20 s for each trial. 2-3 min for resting. Concentrate on target located 2 m away.	Force plate	100	CoP excursion	FFA > SPU
Lenka and Tiberwala (2010)	TTA	Stand still with footwear and hands placed alongside. Toe angle is 30˚. Feet apart for 6cm. Duration for 2 min. Resting for 1min.	Two load cell based unidirectional force platform	260	CoP excursion Stump length RMS	CoP in EC > CoP in EO Sway area in SS > MS SS = MS
Duclos et al. (2009)	TFA, TTA, KD, control	Stand still with feet 3 cm apart from heels. 2 trials performed and 1 min for each trial. Resting for 2 min.	NeuroCoM Balance Master®	100	CoP excursion	TTA : NA leg > A leg; Control : slightly < in dominant leg (in both EO and EC)
					CoP velocity	Amp > control in both EO and EC
Rougier and Bergeau (2009)	TFA, TTA	Stand still with barefoot and hands placed alongside. Toe angle is 30˚. Feet separated 9 cm apart. 5 trials performed and 32 s for each trial. Resting for 32 s.	Equi+ force platform	64	CoP excursion	AP and ML shifted in TFA > TTA TTA: CoP shifted forward in NA leg
					CoP amplitude	NA leg > A leg in AP and ML for TFA and TTA
					RMS	CoP: AP = ML

Table 2.1, continued: Studies reporting balance control in lower-extremity amputees.

Study	Groups	Protocol	Instrumentation	Sampling rate (Hz)	Outcomes measures	Findings
Vanicek, Strike, McNaughton, and Polman (2009)	TTA, control	Stand still with hands placed alongside. 3 trials performed and 20 s for each trial.	NeuroCom Equitest®	100	SOT MCT	Faller = Nonfaller Faller = Nonfaller
Vrieling et al. (2008)	TFA, TTA, control	Hands placed alongside. Feet position was self-selected. Duration for 60 s. 1 min for resting. The platform movements were moved sinusoidally at 1 Hz.	Computer assisted rehabilitation environment	100	GRFy	NA leg: Amp > control A leg : Amp > control in EO
					CoPy excursion	NA leg: Amp > control in EO A leg : Amp < control in EC
Duclos et al. (2007)	TFA, TTA, KD, control	Stand still with feet separated 3 cm apart from heels. Duration for 1 min. 5 min for resting. 2 pre-vibration and 7 post-vibration tests.	NeuroCom Equitest®	100	CoP excursion	NA leg: Amp > control in EO A leg : Amp < control in EC
					CoP velocity	Amp > control
					CoP excursion in muscle vibration effect	Trapezius vibration: Sway in amp > control Gluteus medius vibration: Sway in amp = control
Quai, Brauer, and Nitz (2005)	TTA	Stand still with heel separate 17 cm apart and toe angle is 14˚. Duration for 40 s.	Kistler force plate	100	CoP excursion	EO: NA > A leg in AP and ML EC: NA > A leg in AP and ML
					CoP excursion in Vibration	EO&EC: AP > ML in NA leg
					CoP excursion in pulse score	Lower scorer have greater CoP in ML
Buckley, O'Driscoll, and Bennett (2002)	TFA, TTA, control	Intervention for 3 times per week with 30 min per session. Comfortable standing with hands placed on hips. Feet placed greater than 15 cm. 3 trials performed and 30 s for each trial.	Kistler force plate, modified dynamic stabilimeter	100	CoP excursion	AP > ML AP and ML: Amp > control
					BD	Amp < control in both AP and ML Balance in EO > EC

Table 2.1, continued: Studies reporting balance control in lower-extremity amputees.

Study	Groups	Protocol	Instrumentation	Sampling rate (Hz)	Outcomes measures	Findings
Nadollek, Brauer, and Isles (2002)	TTA	Stand still with feet placed separately for 17 cm between heels. Toe angle is 14˚. Duration for 40 s.	Kistler force plate, dynamometer, B&L stride analyzer	100	CoP excursion	NA leg > A leg in EO and EC AP : EO < EC, ML: EO = EC
Viton et al. (2000)	TTA, control	Stand still with feet placed 10 cm apart for 20 trials. Laterally raised one leg as fast as possible to 45˚.	Kistler force plate	100	CoG position RMS	AP: Amp > control ML: Amp = control Control > Amp

A leg: amputated leg; Amp: amputee; AP: anterior-posterior direction; BD: balance in dynamic standing; BPS: Bipedal Standing; BS: balance in static standing; DCL: Directional control; EC: eyes closed; EO: eyes open; FFA: first-fitted amputee; MCT: motor control test; ML: medial-lateral direction; MS: medium stump; MVL: maximum velocity; MXE: maximum excursion; NA leg: non-amputated leg; LOS: Limits of Stability Test; RMS: root mean square; SOT: Sensory Organization Test; SPU: skilled user; SS: short stump; TFA: transfemoral amputee; TTA: transtibial amputee; UPS: unipedal standing; VN: Van Ness amputee

2.4.2 Variability in Measured Parameter

The most frequently measured parameter is CoP sway. The CoP-related outcomes measured were CoP and center of gravity (CoG) excursion (Table 2.2), CoP velocity (Table 2.3), CoP amplitude (Table 2.4), and root mean square of CoP (Table 2.5). The majority of the studies used these CoP-related outcomes (Andrysek et al., 2012; Buckley, O'Driscoll, & Bennett, 2002; Clark & Zernicke, 1981; Duclos et al., 2009; Duclos et al., 2007; Fernie & Holliday, 1978; Hermodsson et al., 1994; Hlavackova et al., 2011; Lenka & Tiberwala, 2010; Mayer et al., 2011; Nadollek, Brauer, & Isles, 2002; Quai, Brauer, & Nitz, 2005; Rougier & Bergeau, 2009; Vittas, Larsen, & Jansen, 1986; Vrieling et al., 2008). These studies investigated the association between CoP displacement and other subscales: load distribution, level of amputation, duration of wearing prostheses, sensory perception, and sway direction. The studies of Curtze et al. (2012) and Gaunaurd et al. (2011) evaluated ankle and hip moment, while Gaunaurd et al. (2011) and Lenka and Tiberwala (2010) examined stump length as a variable in their studies. With the CoP-related variables, the changes of CoP in balance performance were examined and served as evidence for the association with other subscales.

Overall, the evidence indicates that LEA patients exhibit greater postural imbalance compared with healthy controls. Duclos et al. (2009) found that the mean CoP position of LEA patients generally veered toward the non-amputated (NA) side (7 ± 2 mm), whereas in the healthy controls, it was only slightly toward the dominant side (2 ± 1 mm). Higher CoP velocities were found in LEA patients than in healthy controls (Duclos et al., 2009; Duclos et al., 2007; Fernie & Holliday, 1978). Additionally, the NA side also showed greater amplitudes of CoP along the AP and ML directions (Hlavackova et al., 2011; Rougier & Bergeau, 2009).

Table 2.2: Summary of studies on the CoP and CoG excursion for eyes-open and eyes closed condition.

Study	Visual condition	Parameter	Amputee	Control
Mayer et al. (2011)	EO	D_{CoP} (AP)	SPU : 331.1 ± 151.4 mm	–
			FFA : 398.9 ± 141.1 mm	–
		D_{CoP} (ML)	SPU : 255.9 ± 139.6 mm	–
			FFA : 378.7 ± 154.5 mm	–
Duclos et al. (2009)	EO	D_{CoP}	7 ± 2 mm	2 ± 1 mm
	EC	D_{CoP}	11 ± 3 mm	2 ± 1 mm
Duclos et al. (2007)	EC	D_{CoP}	12.5 ± 4.0 mm	1 ± 2 mm
Quai, Brauer, and Nitz (2005)	EO	D_{CoP} (AP)	NA : 84.0 ± 41.6 mm	–
			A : 43.4 ± 23.8 mm	–
		D_{CoP} (ML)	NA : 15.2 ± 11.7 mm	–
			A : 15.1 ± 7.7 mm	–
	EC	D_{CoP} (AP)	NA : 119.0 ± 49.8 mm	–
			A : 52.8 ± 20.5 mm	–
		D_{CoP} (ML)	NA : 20.2 ± 13.8 mm	–
			A : 19.2 ± 9.9 mm	–
Buckley, O'Driscoll, and Bennett (2002) †	–	D_{CoP}	AP : ~12 mm	AP : ~5 mm
			ML : ~15 mm	ML : ~10mm
Viton et al. (2000)	EO	D_{CoG}	AP : 12.8 ± 2.7 mm	–
			ML : 6.97 ± 1.04 mm	ML : 2.4 ± 0.76 mm
Mouchnino et al. (1998)	EO	D_{CoG}	AP : 12.8 ± 2.7 mm	–
			ML : 6.97 ± 1.04 mm	ML : 2.4 ± 0.76 mm
Clark and Zernicke (1981)	EO	D_{CoP}	37 ± 14 mm	–

A leg: amputated leg; AP: anterior-posterior direction; D_{CoP}: mean of CoP displacement; EC: eyes closed; EO: eyes open; FFA: first-fitted amputee; ML: medial-lateral direction; NA leg: non-amputated leg; SPU: skilled user; † Data extracted from bar chart. All values are reported as mean ± SD.

Table 2.3: Summary of studies on the CoP velocity for eyes-open and eyes closed condition.

Study	Visual condition	Parameter	Amputee	Control
Andrysek et al. (2012)	EO	V_{CoP} (AP)	776.0 ± 359.5 mm/s	494.5 ± 190.5 mm/s
		V_{CoP} (ML)	588.5 ± 328.2 mm/s	241.0 ± 90 mm/s
Hlavackova et al. (2011) †	EC	V_{CoP}	NA : ~35 mm/s	–
			A : ~15 mm/s	–
Mayer et al. (2011)	EO	V_{CoP}	SPU : 23.4 ± 8.3 mm/s	–
			FFA : 33.1 ± 13.4 mm/s	–
Lenka and Tiberwala (2010)	EO	V_{CoP} (AP)	SS : 1.34 ± 0.17 mm/s	–
			MS : 1.21 ± 0.18 mm/s	–
		V_{CoP} (ML)	SS : 2.63 ± 0.33 mm/s	–
			MS : 2.42 ± 0.31 mm/s	–
Duclos et al. (2009)	EO	V_{CoP} †	~ 3.0 mm/s	~ 4.0 mm/s
	EC	V_{CoP} †	~ 3.0 mm/s	~ 3.5 mm/s

A leg: amputated leg; AP: anterior-posterior direction; EC: eyes closed; EO: eyes open; ML: medial-lateral direction; FFA: first-fitted amputee; MS: medium stump; NA leg: non-amputated leg; SPU: skilled user; SS: short stump; TFA: transfemoral amputee; TTA: transtibial amputee; V_{CoP} : mean velocity; † Data extracted from bar chart. All values are reported as mean ± SD.

Table 2.3, continued: Summary of studies on the CoP velocity for eyes-open and eyes closed condition.

Study	Visual condition	Parameter	Amputee	Control
Duclos et al. (2007)	EC	V_{CoP}	15.0 ± 2 mm/s	10.0 ± 1 mm/s
Fernie and Holliday (1978)	EO	V_{CoP}	TFA: 0.91 mm/s TTA: 1.17 mm/s	0.97 mm/s
	EC	V_{CoP}	TFA: 1.80 mm/s TTA: 2.14 mm/s	1.39 mm/s

A leg: amputated leg; AP: anterior-posterior direction; EC: eyes closed; EO: eyes open; ML: medial-lateral direction; FFA: first-fitted amputee; MS: medium stump; NA leg: non-amputated leg; SPU: skilled user; SS: short stump; TFA: transfemoral amputee; TTA: transtibial amputee; V_{CoP} : mean velocity; † Data extracted from bar chart. All values are reported as mean ± SD.

Table 2.4: Summary of studies on CoP amplitude and sway area for eyes-open and eyes closed condition.

Study	Visual condition	Parameter	Amputee	Control
Andrysek et al. (2012)	EO	Sway Area	1493 ± 1420 mm^2	364 ± 200 mm^2
Hlavackova et al. (2011) †	EC	A_{CoP}	NA : ~ 8.2 mm A : ~ 3.5 mm	– –
		Reg_{CoP}	NA : ~ 0.7 A : ~ 0.9	– –
Lenka and Tiberwala (2010)	EO	Sway Area	SS : 93.84 ± 17.47 mm^2 MS : 50.6 ± 22.60 mm^2	– –

A leg: amputated leg; A_{CoP}: amplitude of CoP; EC: eyes closed; EO: eyes open; MS: medium stump; NA leg: non-amputated leg; Reg_{CoP}: regularity of CoP; SS: short stump; † Data extracted from bar chart. All values are reported as mean ± SD.

Table 2.5: Summary of studies on the root mean square (RMS) of CoP for eyes-open and eyes closed condition.

Study	Visual condition	Parameter	Amputee	Control
Andrysek et al. (2012)	EO	*RMS* (AP) *RMS* (ML)	8.1 ± 3.7 mm 10.3 ± 5.3 mm	5.1 ± 2.0 mm 5.7 ± 1.2 mm
	EO	*RMS* (AP)	SS : 1.66 ± 1.04 mm MS : 1.70 ± 0.5 mm	– –
		RMS (ML)	SS : 2.02 ±1.02 mm MS : 1.74 ± 0.5 mm	– –
Vrieling et al. (2008)‡	EO	*RMS*	NA : 3.38 ± 1.69 mm A : 1.36 ± 0.41 mm	1.91 ± 0.62 mm
	EC	*RMS*	NA : 4.28 ± 2.18 mm A : 1.39 ± 0.41 mm	2.82 ± 0.87 mm

A leg: amputated leg; AP: anterior-posterior direction; EC: eyes closed; EO: eyes open; ML: medial-lateral direction; MS: medium stump; NA leg: non-amputated leg; RMS: Root mean square; SS: short stump. All values are reported as mean ± SD.

Table 2.5, continued: Summary of studies on the root mean square (RMS) of CoP for eyes-open and eyes closed condition.

Study	Visual condition	Parameter	Amputee	Control
Nadollek, Brauer, and Isles (2002)	EO	*RMS*(AP)	NA : 8.40 ± 4.16 mm	–
			A : 4.34 ± 2.39 mm	–
		RMS(ML)	NA : 1.53 ± 1.16 mm	–
			A : 1.51 ± 0.77 mm	–
	EC	*RMS*(AP)	NA : 11.85 ± 4.98 mm	–
			A : 5.28 ± 2.05 mm	–
		RMS(ML)	NA : 2.02 ± 1.38 mm	–
			A : 1.92 ± 0.99 mm	–
Viton et al. (2000)	EO	*RMS*	4.69 ± 2.55 mm	5.54 ± 2.78 mm
Mouchnino et al. (1998)	EO	*RMS*	4.69 ± 2.55 mm	5.54 ± 2.78 mm
Clark and Zernicke (1981)	EO	*RMS*(AP)	6.1 ± 3.2 mm	–
		RMS (ML)	3.5 ± 0.6 mm	–

A leg: amputated leg; AP: anterior-posterior direction; EC: eyes closed; EO: eyes open; ML: medial-lateral direction; MS: medium stump; NA leg: non-amputated leg; RMS: Root mean square; SS: short stump. All values are reported as mean ± SD.

2.4.3 Load Distribution

Load distribution asymmetry from the aetiology adaptation process, with postural sway concentrated under the intact leg, may also increase the postural sway. The findings of Nadollek, Brauer, and Isles (2002) and Nederhand et al. (2012) showed that the weight distribution of intact leg was larger compared with the amputated leg. Load distribution was generally asymmetrical among LEA patients (Figure 2.144). The load distribution asymmetry for first-fitted users was greater than skilled prosthetic users (Mayer et al., 2011), whereas the healthy controls had equal distribution between both legs. In Curtze et al. (2012), 69% of TTA's body weight was distributed more under the ankle of the NA leg, whereas healthy controls has equally distributed load under the ankle joints (left: 42%; right: 36%) and hip joints (left: 12%; right: 11%).

Table 2.6 showed the recent findings of standing balance in TTA. By considering the pattern of load distribution, it may augment the reliability of CoP sway in predicting balance performance. The insecure attachment of prosthesis may also affect the LEA's emotions, and weight bearing in the lower extremity (Shurr & Cook, 1990). A continuous period of loading body weight at the NA side may cause knee pain, hip pain, back pain, and knee or hip osteoarthritis at the intact limb (Gailey, Allen, Castles, Kucharik, & Roeder, 2008).

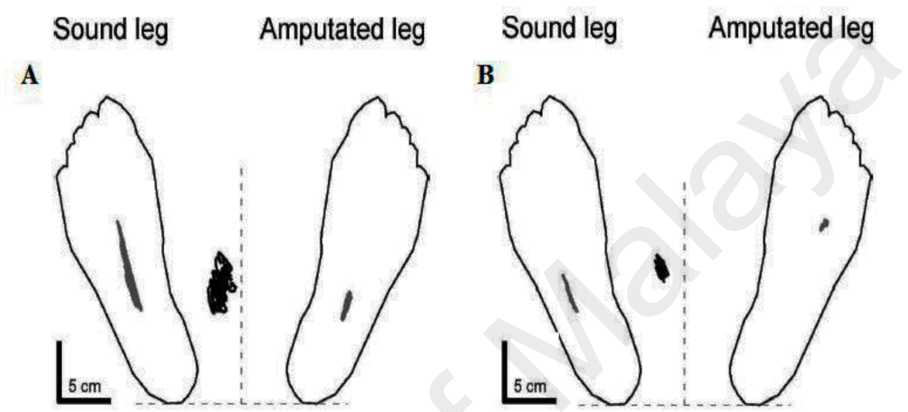

Figure 2.14: The asymmetrical body weight distribution with the resultant CoP trajectories in (A) transfemoral amputee and (B) transtibial amputee. (Adapted from Rougier and Bergeau (2009))

Table 2.6: Summary of studies on the load distribution.

Study	Groups	Years since amputation	Findings
Nederhand et al. (2012)	TFA	10.3 ± 3.9	NA leg > A leg
	TTA	11.8 ± 10.0	
Hlavackova et al. (2011)	TFA	5.8 ± 2.5	NA leg > A leg
Mayer et al. (2011)	TTA-SPU	4.15 ± 2.4	FFA > SPU
	TTA-FFA	0.47	
Duclos et al. (2009)	TFA	8.3 ± 8.6	Amp > control in EO & EC;
	TTA	4.2 ± 3.9	Asymmetry load in EO < EC
	KD	4.5	
	Control	N	
Vrieling et al. (2008)	TFA TTA	21.5	Asymmetry : Amp > control
	Control	N	
Rougier and Bergeau (2009)	TFA	-	Asymmetry: TFA > TTA
	TTA	-	
Nadollek, Brauer, and Isles (2002)	TTA	2.98	NA leg > A leg Performance in EO = EC

All values are reported as mean ± SD.

2.4.4 Sensory Perceptive

Due to the amputation, sensory information will be increased in order to compensate for the lost body segment. Sensory perception can be categorized into visual and proprioception inputs. There are studies investigated CoP changes under different visual conditions (open/closed eyes) (Buckley, O'Driscoll, & Bennett, 2002; Duclos et al., 2009; Fernie & Holliday, 1978; Hermodsson et al., 1994; Lenka & Tiberwala, 2010; Nadollek, Brauer, & Isles, 2002; Quai, Brauer, & Nitz, 2005; Vanicek et al., 2009; Vrieling et al., 2008). Visual input is essential to compensate for the balance impairment due to amputation (Table 2.2 – Table 2.5).

The absence of visual input may increase postural sway among LEA patients. The stance asymmetry may also affected by vision (Buckley, O'Driscoll, & Bennett, 2002; Duclos et al., 2009; Lenka & Tiberwala, 2010; Nadollek, Brauer, & Isles, 2002). AP sway fluctuated more under the absence of visual input than with visual. Nadollek, Brauer, and Isles (2002) highlighted that an increase in AP sway displacement occurs under the NA leg for the closed eyes condition compared with the open eyes condition, whereas no ML sway difference between both legs was observed under either condition. In contrast, Vrieling et al. (2008) found no difference, suggesting that visual input does not affect postural control, as CoP sway is similar in both conditions. The proprioception inputs is dependent upon the intact leg and it may limit sensory feedback. The magnitude of proprioception input can also be affected by visual input and muscle vibration (Barnett, Vanicek, & Polman, 2012; Duclos et al., 2007; Vanicek et al., 2009). The effect of proprioception input (Barnett, Vanicek, & Polman, 2012; Quai, Brauer, & Nitz, 2005; Vanicek et al., 2009) and muscle vibration (Aruin, Nicholas, & Latash, 1997; Duclos et al., 2007) toward changes in balance stability was also studied. Only a few studies investigated these subscales. Balance performance is significantly increased with the aid of visual and proprioception inputs (Barnett, Vanicek, & Polman, 2012). Therefore, sensory system may exhibit a distinct impact on balance, and the compensation mechanism also depends on the change of input information from the sensory system as well as the level of amputation.

2.4.5 Other Variables

Since only a limited number of studies examined on the association between stump length and balance ability, it does not provide a solid evidence that the stump length would affect stability. The sway area for individuals with short stump was larger than individuals with medium stumps (Lenka & Tiberwala, 2010). In Gaunaurd et al. (2011), limb length discrepancy was found in 66% of the LEA participants. This study showed the effect of limb length on the balance stability of LEA patients. The failures of studies to report on the lifestyle of LEA patients and the time of day for research were common deficiencies. The level of physical activity related to muscle strength and the time of day during experiments should be considered as potential factors, as the impact on postural control showed significance in quiet standing.

Given the intricate interaction among balance domains, the understanding in measured balance variables is crucial for improving postural control and helps in daily activity, rehabilitation, psychological functioning and employment management. In terms of postural control and reason for amputation, limited studies reported their association since the related details collected in the methodology were insufficient. Given the limited information collected, some possible balance-related factors may still be absent in the current findings. The effect of stump length and type of prosthetic may play a role as balance-related factors. The justification associated with postural sway is often very important. As no test regarding confidence level (fear of fall) and sleep quality have been conducted, although these may be the behavioral factors for balance, these factors should be tested in future research. The age effect will increase the postural sway for the elderly in the normal population, and this should be considered during experiments. A broad range of age groups throughout the studies limited the analysis of measured variables within particular age groups. The gender effect should also be taken into consideration.

The subsequent chapter describes in detail the study design, recruitment process, subject preparation and data analyses for this research. The research was experimental and focused mainly on laboratory testing. There are five studies performed in this research. Since most of the studies were often interested in qualifying the physical function of the prosthetic (Kegel, Webster, & Burgess, 1980; Lin, Winston, Mitchell, Girlinghouse, & Crochet, 2014; Ventura, Klute, & Neptune, 2011), two studies (Chapter 3 and Chapter 4) are specifically focused on the postural stability of healthy young population. Due to the research using similar equipment, some of the content might be repeated. The study in Chapter 3 evaluated the postural stability of healthy individual in erect posture. The study in Chapter 4 examined the static postural stability under voluntary stance with a toe-extension condition. The studies for Chapter 5 and Chapter 6 investigated the postural stability and muscular response under static and dynamic erect posture among middle-aged adult and TTAs. Chapter 7 evaluated the possible balance-related factors of postural stability among TTA. The detailed methods and techniques of each clinical experiment are described in the following sections. Due to the experiment protocol, some techniques and instruments will be explained and replicated for relevant sections.

CHAPTER 3:

BIOMECHANICAL EVALUATION OF THE RELATIONSHIP BETWEEN POSTURAL CONTROL AND BODY MASS INDEX

3.1 Introduction

In recent years, obesity has rapidly become a global problem and the number of obese individuals is gradually increasing every year throughout the world. If there is a relationship between the weight of an individual and their ability to balance, this could have severe implications, as poor balance is considered to be one of the major risk factors for the occurrence of falls that could lead to severe injury or death.

An individual's postural control involves a complex system that allows them to maintain balance during quiet standing. The information that is required in order to sustain balance is secured from the physiological system which includes the vestibular, proprioception and visual systems (Karimi et al., 2008). Shumway-Cook and Woollacott (2007) described that there is small amount of spontaneous postural sway in quiet standing. Aside from that, researchers have found that the responses of the multiple body segments may affect postural control (Hodges, Gurfinkel, Brumagne, Smith, & Cordo, 2002). During the balance against perturbation in quiet standing, ankle, hip and stepping strategies were used as the movement patterns that recover stability by regulating the CoM in the sagittal plane. According to Abe, Masani, Nozaki, Akai, and Nakazawa (2010), the acceleration of CoM in the medial-lateral direction is a more plausible determinant that is induced by the musculoskeletal system. Nevertheless, Granacher and Gollhofer (2011) suggested that no significant interaction occur between variable of postural control and muscle strength, but, significant interaction has been shown between the variables of isometric and dynamic muscle strength.

Obesity is generally acknowledged as an excess of body mass and body adipose tissue distribution that could increase health risks and decrease of occupational and recreational comfort as a result of injuries such as sprains, strains and dislocations (Matter et al., 2007). Poor postural performance and reduced motor activity were shown to be more prevalent among obese children than in non-obese children (Matter et al., 2007). Fjeldstad, Fjeldstad, Acree, Nickel, and Gardner (2008) also documented that obese elderly groups experienced a high prevalence of fall, poor health and a lower quality of life and claimed that adipose tissue distribution could be considered as a major contributory factor to balance impairment. Increased obesity had been proven to have an impact on postural control in terms of postural sway, energy cost, attentional cost, motor reaction time and muscular torque (Katch, Becque, Marks, Moorehead, & Rocchini, 1988; Salsabili et al., 2011).

The clinical tools that are commonly used for the assessment of balance performance are the force plate, Lord sway-meter and star excursion balance test (Blaszczyk et al., 2009; Colné, Frelut, Pérès, & Thoumie, 2008). However, in recent years, the Biodex Balance System SD provides distinct advantage in data measurement such that various types of test are available and convenient for data error checking. It is easy to administer, available for multiples range of people and simple to interpret (Aydoğ, Bal, Aydoğ, & Çakei, 2006; Salsabili et al., 2011). The study aim in this chapter was to assess the postural activity among healthy young adults according to their BMI classification (underweight/ normal weight/ overweight/ obese) and gender during two types of quiet standing condition (bipedic stance and unipedic stance) in order to elucidate whether BMI does act as an indicator of an individual’s ability to sustain postural ability. It was hypothesized that a difference in postural activity may be caused by an individual’s BMI whether they were male or female.

3.2 Methods

3.2.1 Participants

Forty healthy male subjects (age =22.1 ± 2.0 years; height=1.71 ± 0.07 m; mass=73.4 ± 20.5 kg) and forty female subjects (age =21.4 ± 1.8 years; height=1.58 ± 0.05 m; mass=61.6 ±15.5 kg) consented to participate in the study. The subject inclusion criteria consisted of subjects' age (between 19 and 26 years old) and in good health. None of the subjects had undergone previous balance training programs using the Biodex Balance System SD (BBS) prior to the study, and none of them had suffered from any prior lower limb injury, neurological, vestibular impairment, or balance disorders. The research was implemented in a cross-over study design, where all subjects randomly underwent the same procedures. Approval from the Institutional Review Board for the test procedure was obtained.

3.2.2 Instrumentations

Subjects were categorized into four groups based on their BMI: underweight, normal weight, overweight and obese. In this study, the BBS (Biodex Medical System Inc., Shirley, NY, USA) is utilized to measure the displacement of the CoP and it is composed of a circular platform that allows up to 20° of platform tilt in a 360° range of motion at a sampling rate of 20 Hz. The measures of postural stability score for the BBS are Overall Stability Index (OSI), Medial-Lateral Stability Index (MLSI) and Anterior-Posterior Stability Index (APSI). These indices are standard deviations that assess the path of sway around the zero point from the center of the platform and they are measured in degrees. The stability indices scores show the foot displacement (°) for motion in the sagittal and frontal planes. Within this study, the displacement from horizontal along ML axes as x-direction, and from vertical along AP axes as y-direction were evaluated as MLSI and

APSI respectively. The equations for OSI, MLSI and APSI scores are presented in Equation 3.1 – 3.3:

$$\mathrm{OSI} = \sqrt{\frac{\sum(0-Y)^2 + \sum(0-X)^2}{\mathrm{samples}}} \tag{3.1}$$

$$\mathrm{MLSI} = \sqrt{\frac{\sum(0-X)^2}{\mathrm{samples}}} \tag{3.2}$$

$$\mathrm{APSI} = \sqrt{\frac{\sum(0-Y)^2}{\mathrm{samples}}} \tag{3.3}$$

The BBS recorded the foot displacement in the x-direction and y-direction. Then, the system generated the OSI, APSI and MLSI using the equations above. The OSI score was established by combining the degree of tilt for the AP and ML axes, as this had been suggested as the best balance indicator to measure overall platform balance (Arnold & Schmitz, 1998a). Calibration of the platform actuator and tilt sensor were selected in the BBS before commencing the test. Arnold and Schmitz (1998b) found the intratester reliability of the BBS was 0.82 (OSI), 0.43 (MLSI) and 0.80 (APSI). Cachupe, Shifflett, Kahanov, and Wughalter (2001) showed that the examination of measures across 5 test evaluation indicated that the BBS produced reliable measures at 0.92 (OSI), 0.90 (MLSI) and 0.86 (APSI).

3.2.3 Study Protocol

Subjects stood on the BBS without footwear for all balance tests. The bipedic stance test (BLS) and unipedic stance test (ULS) were accomplished by using BBS to measure the postural balance score under the static level. Test order was randomized during both sets of tests. During the BLS test, subjects were asked to stand still on the platform of the BBS with their arms crossed over their chest. They were required to stand comfortably and maintain their visual level by focusing straight ahead on the monitor. The platform was unlocked and subject was allowed to adjust their foot placement until a comfortable standing position was achieved while they simultaneously maintained a moving pointer at the center point of the monitor. Following this, the platform was locked and foot placement of the subject remained constant throughout the static balance test. Only one practice trial was performed for instrument acquaintance with the BBS to ensure that the collected data was reliable across the spectrum of stability levels. All differences that affected the data collection are not related to learning and the learning effect could be minimized (Pincivero, Lephart, & Henry, 1995). Subjects were encouraged to maintain the moving pointer at the center point throughout the test.

The BLS was assessed at the static level for 30 seconds. Five test evaluations were performed with a rest period of 10 s between each test and the data was averaged. The test protocol for ULS was similar to that utilized during BLS, excluding the standing posture. In ULS, subjects were instructed to stand with the dominant leg, which is defined as the preferential use in voluntary motor acts using ipsilateral members of the lower limb (Lanshammar & Ribom, 2011). The knee of the contra-lateral limb was held in a slightly flexed position to 90°. However, the non-supported limb was not allowed to come into contact with the supporting limb throughout the test. The use of the holding rail was only permitted as a means of avoiding falling. The test was to be aborted and repeated in the event that subject lost balance and fell down.

3.2.4 Statistical Analysis

All the data was presented as mean values and standard deviations (mean ± SD). Dependent-sample *t*-tests were used to compare the postural stability index score between BLS and ULS. Levene's test was used to analyze the equality of variances. For the basis of statistical analysis among groups, normality of the data was investigated using the Kolmogorov-Smirnov test. A two-way analysis of variance (ANOVA) was then performed to determine the main effect of BMI and gender on bipedic and unipedic conditions. *Post-hoc* analysis was performed using the HSD Tukey test to determine where the significant differences occurred. The alpha level was set at $p<0.05$ for all analyses. The analysis of normality test showed that all data was normally distributed. The data was normalized with respect to body weight and relative stability used in the data analysis in order to avoid the potential misinterpretation of data and to reduce the absolute stability differences (Winter & Maughan, 1991). All statistical analysis was performed using the statistical software SPSS 19.0 (Version 16, IBM Corp., Armonk, NY).

3.3 Results

The demographic characteristics of subjects for this study are summarized in Table 3.1. Due to weight-based group selection, there were significant differences in body weight and BMI. The dependent sample *t*-test revealed that there was a significant difference in postural performance between BLS and ULS. The changes in OSI, APSI and MLSI stability scores across the different category classifications of BMI between male and female groups are presented in Table 3.2. The changes in OSI score between BLS and ULS in male and female while using BBS are summarized in Figure 3.1.

Table 3.1: Subjects' (*N*=80) descriptive and demographic characteristic.

Variables	Underweight	Normal weight	Overweight	Obese
Sex ratio (male/female)	10 / 10	10 /10	10 / 10	10 / 10
Age* (years)	21.5 ± 1.5	21.3 ± 1.3	22.4 ± 2.6	21.9 ± 1.8
Height* (m)	1.66 ± 0.09	1.64 ± 0.08	1.64 ± 0.09	1.64 ± 0.10
Weight*, † (kg)	48.2 ± 4.8	57.0 ± 7.7	73.7 ± 9.5	91.2 ± 13.5
BMI *, † (kg/m^2)	17.4 ± 0.9	21.1 ± 1.7	27.4 ± 1.6	33.8 ± 2.4

* Values for age, height, weight are means ± standard deviation
† Significant difference between group (p<0.05)

The result of the two-way ANOVA exhibited significant main effect of BMI on the OSI score ($F_{3,72} = 7.2$; $p < 0.001$), APSI score ($F_{3,72} = 8.0$; $p < 0.001$) and MLSI score ($F_{3,72} = 4.8$; $p = 0.004$). There was significant main effect of gender for BLS is shown in the OSI score ($F_{1,72} = 10.8$; $p = 0.002$), APSI score ($F_{1,72} = 6.2$; $p = 0.015$) and MLSI score ($F_{1,72} = 4.8$; $p = 0.004$). The obese group demonstrated a significant lower mean stability score than the underweight group and normal weight group.

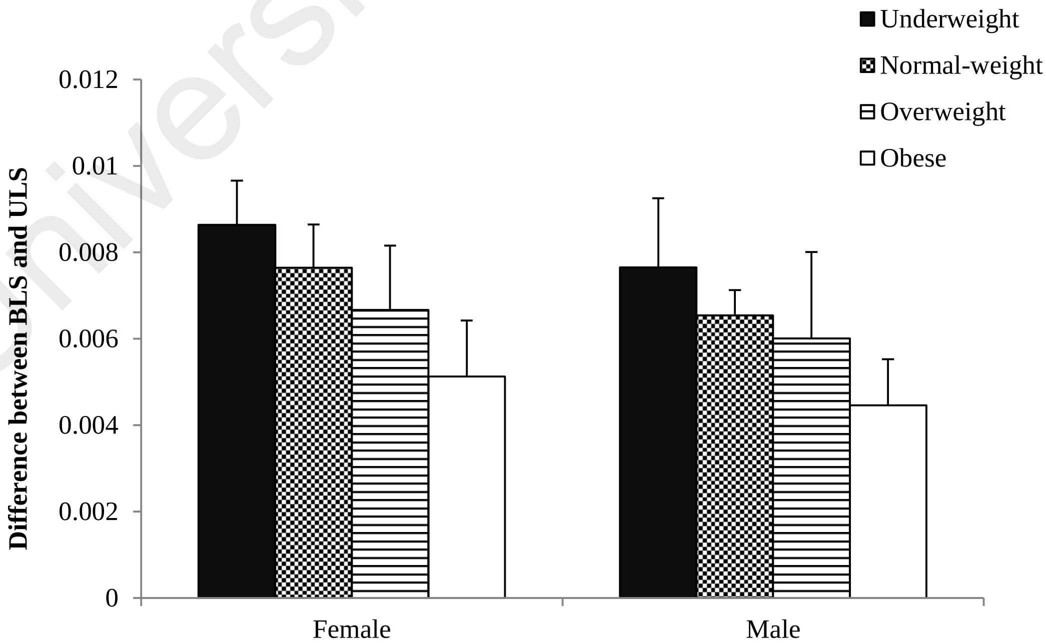

Figure 3.1: The difference of relative stability scores between bipedic stance (BLS) and unipedic stance (ULS) for males and females in mean (± standard error) according to the BMI classification.

In ULS, analysis of variance showed that significant main effect of BMI was found in the OSI score ($F_{3,72} = 16.7$; $p < 0.001$), APSI score ($F_{3,72} = 12.0$; $p < 0.001$) and MLSI score ($F_{3,72} = 10.9$; $p < 0.001$). Significant main effects of gender were found in the OSI score ($F_{1,72} = 12.5$; $p = 0.001$), APSI score ($F_{1,72} = 9.6$; $p = 0.003$) and MLSI score ($F_{1,72} = 4.1$; $p = 0.047$). In the obese group, the mean stability score was significant lower than the other three BMI groups. Generally, for both the BLS and ULS conditions, there are no interactions found between BMI groups and genders for all variables measured. The mean stability score for females in the obese group was lower than the remaining groups. *Post-hoc* pair-wise comparisons showed that the OSI, APSI and MLSI scores of the obese group were significantly lower than the remaining BMI groups (Table 3.2).

The mean of the OSI score in BLS decreased by 5.5%, 18.3% and 15.3% with respect to the increase in BMI, while the stability score in ULS decreased by 9.60%, 14.2% and 22.3% with respect to the increase in BMI value. There were significant, but low, correlations between the stability score and BMI for bipedic stance ($r = -0.511$, $p < 0.001$) and unipedic stance ($r = -0.650$, $p < 0.001$). Individuals in the underweight group are more likely to demonstrate a higher stability score in both conditions. No subject grasped the holding rail to regain balance during the balance tests.

Table 3.2: Summary of the stability score for both gender group in the bipedic and unipedic stance condition.

	Male (n=10)		Female (n=10)		Overall (n=20)	
	Mean ± SD	95% CI	Mean ± SD	95% CI	Mean ± SD	95% CI
Bipedic stance						
OSI						
Underweight	0.006 ± 0.001	0.005-0.007	0.008 ± 0.002 *	0.006-0.009	0.007 ± 0.002	0.006-0.007
Normal Weight	0.005 ± 0.002	0.004-0.007	0.007 ± 0.002 *	0.005-0.008	0.006 ± 0.002	0.005-0.007
Overweight	0.004 ± 0.001	0.003-0.005	0.006 ± 0.002 [a]	0.004-0.007	0.005 ± 0.002[a]	0.004-0.006
Obese	0.004 ± 0.002	0.003-0.005	0.004 ± 0.001 [a, b]	0.003-0.005	0.004 ± 0.001[a, b]	0.004-0.005
APSI						
Underweight	0.004 ± 0.001	0.003-0.005	0.005 ± 0.002	0.004-0.006	0.005 ± 0.001	0.004-0.005
Normal Weight	0.004 ± 0.001	0.003-0.005	0.005 ± 0.002	0.004-0.006	0.004 ± 0.001	0.004-0.005
Overweight	0.003 ± 0.001	0.003-0.004	0.004 ± 0.001	0.003-0.004	0.003 ± 0.001[a, b]	0.003-0.004
Obese	0.003 ± 0.001 [a]	0.002-0.004	0.003 ± 0.001 [a, b]	0.003-0.004	0.003 ± 0.001[a, b]	0.002-0.004
MLSI						
Underweight	0.003 ± 0.002	0.002-0.004	0.004 ± 0.002	0.003-0.005	0.004 ± 0.002	0.003-0.004
Normal Weight	0.002 ± 0.001	0.001-0.003	0.004 ± 0.001 *	0.003-0.005	0.003 ± 0.002	0.002-0.004
Overweight	0.002 ± 0.001	0.001-0.003	0.002 ± 0.002	0.001-0.004	0.002 ± 0.001[a]	0.002-0.003
Obese	0.002 ± 0.001	0.001-0.002	0.002 ± 0.001 [a, b]	0.002-0.003	0.002 ± 0.001[a, b]	0.001-0.002

† $p<0.05$: significant difference in comparison to bilateral stance and unilateral stance.
* $p<0.05$: significant difference in gender group among the same postural condition.
[a] $p<0.05$: significant difference when compare with BMI group (underweight) among the same postural condition by using *Post-hoc* pair-wise comparisons.
[b] $p<0.05$: significant difference when compare with BMI group (normal weight) among the same postural condition by using *Post-hoc* pair-wise comparisons.
[c] $p<0.05$: significant difference when compare with BMI group (overweight) among the same postural condition by using *Post-hoc* pair-wise comparisons.

Figure 3.2, continued: Summary of the stability score for both gender group in the bipedic and unipedic stance condition.

	Male (n=10)		Female (n=10)		Overall (n=20)	
	Mean ± SD	95% CI	Mean ± SD	95% CI	Mean ± SD	95% CI
Unipedic stance						
OSI						
Underweight	0.013 ± 0.003	0.011-0.015	0.016 ± 0.003 *	0.014-0.018	0.015 ± 0.003 †	0.013-0.016
Normal Weight	0.012 ± 0.002	0.010-0.014	0.015 ± 0.002 *	0.013-0.016	0.013 ± 0.003 †	0.012-0.014
Overweight	0.010 ± 0.003 [a]	0.008-0.013	0.012 ± 0.003 [a]	0.010-0.015	0.011 ± 0.003 [†,a,b]	0.010-0.013
Obese	0.009 ± 0.003 [a, b]	0.007-0.010	0.009 ± 0.003 [a, b]	0.008-0.011	0.009 ± 0.003 [†,a,b,c]	0.008-0.010
APSI						
Underweight	0.008 ± 0.001	0.007-0.009	0.011 ± 0.002 *	0.009-0.012	0.010 ± 0.002 †	0.009-0.010
Normal Weight	0.008 ± 0.002	0.007-0.009	0.010 ± 0.002 *	0.009-0.011	0.009 ± 0.002 †	0.008-0.010
Overweight	0.007 ± 0.003	0.005-0.009	0.008 ± 0.002 [a]	0.007-0.010	0.008 ± 0.003 [†,a,b]	0.006-0.009
Obese	0.006 ± 0.003[a, b]	0.004-0.007	0.006 ± 0.001 [a, b]	0.006-0.007	0.006 ± 0.002 [†,a,b,c]	0.005-0.007
MLSI						
Underweight	0.009 ± 0.002	0.007-0.010	0.009 ± 0.002	0.007-0.009	0.009 ± 0.002 †	0.008-0.010
Normal Weight	0.007 ± 0.002	0.006-0.009	0.008 ± 0.002	0.007-0.009	0.008 ± 0.002 †	0.007-0.009
Overweight	0.006 ± 0.002 [a]	0.005-0.007	0.007 ± 0.002[a]	0.006-0.008	0.006 ± 0.002 [†,a,b]	0.005-0.007
Obese	0.005 ± 0.002 [a]	0.004-0.006	0.006 ± 0.003[a]	0.004-0.008	0.005 ± 0.002 [†,a,b]	0.005-0.007

† p<0.05: significant difference in comparison to bilateral stance and unilateral stance.
* p<0.05: significant difference in gender group among the same postural condition.
[a] p<0.05: significant difference when compare with BMI group (underweight) among the same postural condition by using *Post-hoc* pair-wise comparisons.
[b] p<0.05: significant difference when compare with BMI group (normal weight) among the same postural condition by using *Post-hoc* pair-wise comparisons.
[c] p<0.05: significant difference when compare with BMI group (overweight) among the same postural condition by using *Post-hoc* pair-wise comparisons.

3.4 Discussion

The main aim of this study was to investigate the impact of both the BMI and gender of young adults on postural balance activity. The primary findings revealed that those individuals in the underweight group demonstrated a better balance performance, and that postural activity was negatively affected by increased BMI in both BLS and ULS. The findings also revealed that there is a trend in young female adults to generate greater postural sway in AP and ML direction when compared to the young male adults.

In line with previous studies, it is vital to investigate the biomechanical factors that may influence postural activity. Chiari, Rocchi, and Cappello (2002) reported that height, weight, BoS, maximum foot width and feet opening angle should be taken into account when considering biomechanical factors. Age is not a factor that is associated with postural control for different BMI categories (Cruz-Gómez, Plascencia, Villanueva-Padrón, & Jáuregui-Renaud, 2011). The static control balance results for the obese group during the experiment are more revealing with regards to their ability to maintain postural control. A related study revealed that the obese group could not generate sufficient muscle force to control the displacement of CoM (Colné et al., 2008).

The obese group had greater postural sway for BLS and ULS. Weight distribution may relate to the increase of postural sway. The results of this study are in agreement with previous related studies. Hue et al. (2007) argued that obesity is generally associated with greater balance instability, as increases in the contact area and pressure reduce the sensory information executed from the planter mechanoreceptors in middle-aged adults. Corbeil, Simoneau, Rancourt, Tremblay, and Teasdale (2001) claimed that excessive body weight among obese groups will increase falling risks compared to non-obese groups. It is evidenced that weight loss in obese males will directly improve their postural control (Teasdale et al., 2007). Furthermore, Himes (2000) suggested that a sedentary lifestyle might lead to the muscle mass reduction and arise in weight gain and disease prevalence rate.

In the present study, there is a greater difference in ML than AP stability during BLS and ULS, since postural instability is usually associated with an increase in lateral body sway (Blaszczyk, Orawiec, Duda-Klodowska, & Opala, 2007). Generally, balance control in the ML direction occurs at the hip and trunk of the body while the pelvis generates ML motion in the lateral direction (Shumway-Cook & Woollacott, 2007). When the descending response of a body segment takes place, head movement will occur first, followed by trunk and hip movements. The data revealed that there was a significant increase in ML sway for the obese group in ULS compared to the remaining BMI groups. This is in agreement with McGraw et al. (2000) who also found differences in ML stability between obese and non-obese boys. Furthermore, McClenaghan et al. (1996) found a significant difference for postural balance in ML stability between young adults and elderly group. Menegoni et al. (2009) argued that a decrease in AP stability among elderly individuals in the static position could lead to an increase in motor activity. They also indicated that only AP stability is correlated with body weight. Additionally, weight gain in the BMI group induced a similar effect on AP and ML stability in both gender groups. Although all the studies above used static force platform as the assessment toolkit, there is evidence that BBS measurements are more reliable compare with the data obtained from a static force platform (Hinman, 2000).

The current investigation revealed that there is a trend in female subjects displaying a greater postural sway compared to their male counterparts, however these differences were only significant in some underweight and normal weight groups. The greater postural sway that occurred with the female subjects indicated an increase in CoP displacements towards the limit of BoS. However, this finding is in contrast with earlier studies, which found that males had a greater postural sway than females (Mickle, Munro,

& Steele, 2010). It is suggested that the discrepancy in the findings could be related to the adipose tissue mass distribution of the genders, as the android type normally occurs in males and the gynoid type in females. The adipose tissue distribution for the android type (apple-like body shape) is concentrated in the thorax-abdominal region, while for the gynoid type (pear-like body shape), adipose tissue is usually found around the hip and thigh areas (Clark, 2004). Both males and females can have an android body type (Menegoni et al., 2009). Goodman-Gruen and Barret-Connor (1996) described that obese females tend to accumulate a greater amount of adipose tissue in the lower extremities. Additionally, the arch angle of the foot in females presents a greater ligament laxity. Higher body weight and a flexible longitudinal arch would lead to a greater postural sway (Aurichio, Rebelatto, & Castro, 2011). Readers should take note of whatever differences were revealed between male and female groups in this study.

ULS is a clinical tool that is utilized to measure postural steadiness in a static position (Josson, Seiger, & Hirschfeld, 2004), since it is a posture which requires greater motor control than BLS and it requires a posture balance recovery movement at both the ankle and hip joints in order to create musculature restoration forces. In research that assessed the postural control of 453 women age between 20 and 80, it was found that women in their 70s have difficulty in maintaining postural control for BLS and ULS on either leg (Choy, Brauer, & Nitz, 2003). In this study, most of the subjects preferred their right lower limb (86%) as the dominant leg. Greve et al. (2007) observed that no difference in balance performance was found in the dominant leg and non-dominant leg during ULS.

Although there are studies in existence that have investigated postural balance, the impact of BMI on stability was the main concern of this study. Learning may affect the result which influences the proprioception and vestibular equilibrium. In this study, one practice trial was performed before the test started. Although, it might deviate the focus from BMI and gender effect, however, with one practice trial, the subject will be able to learn and adapt for the test first. Five test evaluations were performed in this study. So, the data collected is standardized and accurate without the influencing factor of the learning. In addition to this, the sample size of the groups involved in current study was small (N = 80), and all participants were from one geographical location; both of these facts may undermine the findings. This study provided normative postural stability data for the balance ability of healthy adolescents (between 19 to 26 years old) in a static balance test. Coaches should consider the impact of gender and BMI in practical fields such as athlete selection for sporting activities that require quiet standing. Further studies should take note of muscle activity and other dynamic components during the balance test.

In conclusion, this study revealed a relationship between static balance ability in young adults and their BMI and gender. The data indicated that balance performance was diminished as the BMI value of the subjects increased and subjects that were categorized in the obese group were more likely to exhibit greater postural sway. This indicates that changes in the BMI may alter an individual's ability to balance, although other biomechanical factors may also affect the performance. In terms of static balance, there is a trend towards females displaying a greater postural sway compared to males in certain BMI classifications.

The study in this chapter is mainly focused on the changes of postural stability with BMI values among young adults. It would provide a baseline data for the balance performance of young adult with sedentary lifestyle. Moreover, the content of this study was reproduced from the study of Ku, Abu Osman, Yusof, and Wan Abas (2012a) with the agreement of all authors. The following chapter will demonstrate the changes of postural stability in the standing posture of BLS and active stance with toe-extension. It is important for identify whether there are any changes of balance performance in a circumstances of actively limited base of support.

CHAPTER 4:
THE EFFECT ON HUMAN BALANCE OF STANDING WITH TOE-EXTENSION

4.1 Introduction

Since balancing is a vital component of life for all human beings, balance control has been examined extensively in a number of studies (Sturnieks, George, & Lord, 2008; Winter, 1995). All movements that are performed under static or dynamic conditions are necessary to maintain a good quality of life, and the individual can achieve body equilibrium and maintain stability during quiet standing and ambulatory activities.

Researchers agreed that the human body will attempt to orientate the CoM, which is a virtual point that is equivalent to the total body mass at which the average of mass distribution for each body segment may be assumed to be concentrated, against perturbation (Hernández et al., 2009). A healthy human being will naturally attempt to return to the CoM within the base of support by subconsciously regulating the body's position. The base of support is the displacement region for the CoP, since the CoP serves as the location point of the average distribution for all the pressure over the ground surface contact area (Gravante, Russo, Pomara, & Ridola, 2003). By measuring the displacement of the CoP, an individual's balance can be assessed. Winter (1995) argued that the location of the CoP is impacted directly by an individual's foot activity. He stated that the CoP will move in a medial direction as a result of an increase of evertor activity, while the CoM shifts laterally, and will move in a lateral direction as a result of an increase in the invertor activity, which results in the medial shift of the CoM (Gefen, Megido-Ravid, Itzchak, & Arcan, 2002).

Numerous studies have investigated the role that the function of the foot plays in standing, heel-toe standing, heel-standing and full toe-standing (Nolan & Kerrigan, 2004).

One of factors in achieving postural balance is the anatomy of the foot. The contribution of the foot mechanism towards the stability of postural standing has attracted attention from various investigators. In particular, it has been proven that the normal arch in the foot plays a role in shock absorption and propulsion phases, whereas the toe is responsible for providing a stable surface area that remains in contact with the ground and serves to relay relevant sensory proprioception information to the central nervous system. This information is then used to perform basic corrections when balance disturbances occurred (Tortolero, Masani, Maluly, & Popovic, 2007). Cavanagh, Rodgers, and Liboshi (1987) reported that 60% of weight-bearing pressure is distributed at the heel of the foot, 8% at the mid-foot, and 28% at the forefoot. With regards to pressure distribution, studies have found that the highest peak pressure at the toes were found at the great toe (30%), followed by the second toe (24%), third toe (21%), fourth toe (16%) and small toe (9%) (Hughes, Clark, & Klenerman, 1990). In addition, the second and third metatarsal areas display the highest pressure at the forefoot region for a normal, healthy individual (Cavanagh, Rodgers, & Liboshi, 1987). According to Hicks (1954), the extension of toes may subsequently lead to the following situations: (i) the elevation of the foot arch, (ii) the supination of the feet occurring at the posterior part of the foot, (iii) a lateral rotation of the leg, (iv) a tightness of the band at the region of the plantar aponeurosis.

In recent studies, the BBS has been more frequently used as a tool for balance assessment than the former force platform. It is a multiaxial device that evaluates and measures the CoP for up to 20° of platform tilt in a 360° range of motion. The BSS can compute three measures: the OSI, MLSI and APSI. These variables were used in this study to measure and evaluate balance performance. The main study purpose in this chapter was to determine how changes in BLS and stance with toe-extension (SWT) conditions impact postural control. Furthermore, the study sought to examine the extent to which these impacts differ according to gender. The researchers hypothesized that the standing postures of BLS and SWT would not impact the postural sway in terms of the OSI, APSI or MLSI.

4.2 Methods

4.2.1 Participants

Thirty healthy young adults (15 male and 15 female) with no prior lower-limb injuries were recruited for this study. None of the participants had experienced previous balance training using BBS, or had suffered from any neurological, vestibular or balance impairments. The research incorporated a crossover study design, where all participants underwent the same test protocols. In order to certify the health condition of the participants with regards to their daily routines and lifestyle, a basic lifestyle survey was undertaken before the study began. Participants who engaged in an active lifestyle and generally wore flat shoes were selected for this study. The study protocol was reviewed and approved by the Medical Ethic Committee in University of Malaya Medical Centre. All participants have signed their written informed consent form to participate in the study.

4.2.2 Instrumentation

The BBS used in this study contained four strain gauges under a circular platform in order to measure the displacement of CoP at a sampling rate of 20 Hz. For the OSI, MLSI and APSI stability scores, these indices were the measures of standard deviation, which were assessed along the path of sway around the zero point from the center of the platform of BBS. The units were recorded in degrees. The foot displacements that occurred on the ML axis were labelled as "x-direction", while those on the AP axis were labelled as "y-direction," and these variables were measured as the MLSI and APSI respectively. In turn, the OSI, which is sensitive to the change in body sway across the ML and AP axes, was derived from the MLSI and APSI.

Arnold and Schmitz (1998a) suggested that the OSI is an important tool that can be used to assess balance stability, since the OSI can be affected by both AP and ML directions. Previous studies have found that the reliability of the BBS for 5 trials was 0.92 (OSI), 0.90 (MLSI) and 0.86 (APSI). Within this study, Biodex Medical Systems software (Version 1.33) was used to compute the transmitted data and generate the stability indices that were used to evaluate the data.

4.2.3 Study Protocol

In the current study, the BBS was used to evaluate the static postural control during (i) BLS (Figure 4.1A), and (ii) SWT (Figure 4.1B). For the static level, the circular platform was set to remain static with no platform tilt allowed. The participants randomly performed the balance tests for both sets of tests. During BLS, participants were instructed to stand barefooted on the BBS platform. Both of their hands were placed over their chests and they were asked to look straight forward. They were then instructed to adjust both their supporting feet in order to achieve a comfortable standing posture that allowed them to maintain balance. The instructions for position adjustment were provided on the instrument panel. Once the participant was in a comfortable standing position, the platform was locked and the foot placement of the participant subsequently remained constant throughout the test. When the test was commenced, participants were required to concentrate on maintaining the moving pointer at the center of the circle on the instrument panel for a period of 30 s. Each participant underwent 5 trials and a rest period of 10 s was provided between each trial to enable the participant to momentarily relax their body without looking at the panel. Five trials were performed continuously and the average score was computed.

Figure 4.1: Experimental setup (A). Illustration for the setup that allow the balance assessment during BLS by using the Biodex Balance System SD. (B).Illustration for the assessment during the stance with toe-extension condition.

The same procedure was applied during the SWT as that utilized in BLS test, excluding the element that involved the standing posture. During the SWT, participants were instructed to lift their toes as high as they could (normally about 90° to the line of metatarsal as shown in Figure 4.2) throughout the test in order to avoid ground contact (Hicks, 1954). Real-time foot monitoring was employed to monitor the angle of the participant’s toes. A verbal reminder was given if their toes dropped below the 5° limit.

If the participant’s toes dropped more than 6°, the trial was cancelled and the test was repeated.

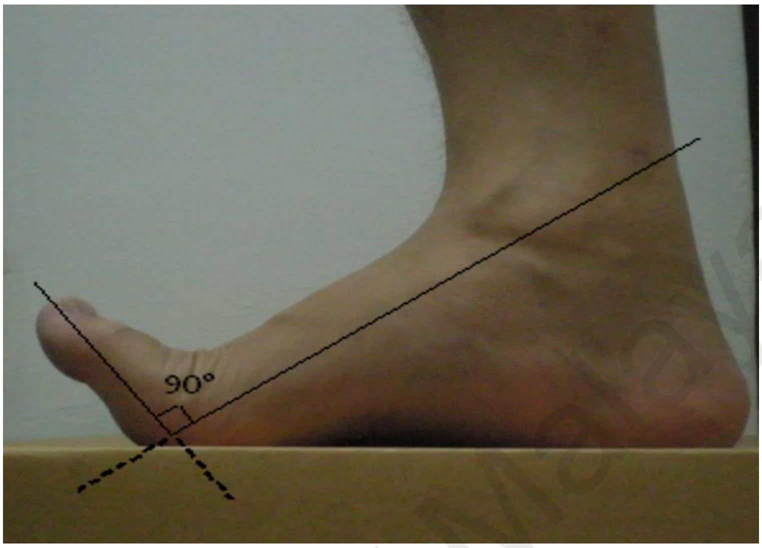

Figure 4.2: Foot posture for stance with toe-extension. Illustration for the balance assessment of stance with toe-extension condition using BBS.

4.2.4 Statistical Analysis

All statistical analysis was performed using SPSS for Windows (Version 19; IBM Corp., Armonk, NY). A 2 × 3 × 2 (standing posture × stability index × gender) mixed factor repeated measures ANOVA was used for the purposes of the analysis. There was no interaction of gender and stability index, hence the groups were collapsed for gender. A 2–way interaction was seen for standing posture and stability index. The Shapiro–Wilk Test was used to assess the normality of the data. The test of normality verified that all the data produced was normally distributed. The data was normalized with respect to body weight, and relative stability was used in the data analysis in order to avoid a potential misinterpretation of data and to reduce the absolute stability differences (Winter

& Maughan, 1991). Relative stability was defined as the ratio of total stability index score over body mass, which was expressed as degree per kilogram.

4.3 Results

The descriptive statistical analysis of the demographic characteristics for all participants is tabulated in Table 4.1. The results of this study elucidated that there was no significant main effect between BLS and SWT in the OSI, APSI and MLSI scores among the 30 participants ($F_{2, 28} = 3.357$, $p = 0.077$). The main effect of balance was significant ($F_{2, 28} = 275.1$, $p < 0.001$). The Bonferroni *post hoc* test showed significant differences between the OSI with MLSI ($p < 0.001$), the OSI with APSI ($p < 0.001$) and the MLSI with APSI ($p < 0.001$). The standing posture × stability index interaction was significant ($F_{2, 28} = 13.64$, $p < 0.001$).

Table 4.1: Descriptive and demographic characteristic for all participants.

Variables	Male	Female	Overall
No. of individual (*n*)	15	15	30
Age (year)	21.2 ± 1.6	21.1 ± 1.0	21.2 ± 1.3
Height * (m)	1.67 ± 0.06	1.59 ± 0.06	1.63 ± 0.07
Mass * (kg)	62.0 ± 8.4	49.9 ± 5.5	56.0 ± 9.3
BMI * (kg/m^2)	22.3 ± 3.2	19.7 ± 2.4	21.0 ± 3.1

* Significantly different between gender ($p < 0.05$).

Considering the available number of participants in this study, the within condition analysis revealed significant differences between the OSI with APSI, the OSI with MLSI, and the APSI with MLSI in BLS (all $p < 0.001$), while differences were shown between the OSI and APSI, and the OSI and MLSI in SWT ($p < 0.001$). The means of the relative stability index for both BLS and SWT conditions are displayed in Table 4.2 – 4.4.

Table 4.2: Descriptive statistics for the balance stability mean score for male.

Conditions	Male		
	Mean ± SD	95% CI	CV (%)
Bipedic stance			
OSI (˚)	0.006 ± 0.001	0.006-0.007	22.9
MLSI (°)	0.002 ± 0.001	0.002-0.003	40.5
APSI (°)	0.005 ± 0.002 [b]	0.004-0.006	35.2
Stance with toe-extension			
OSI (˚)	0.007 ± 0.003	0.006-0.009	40.0
MLSI (°)	0.005 ± 0.003 [a]	0.003-0.006	58.2
APSI (°)	0.005 ± 0.002	0.004-0.006	38.4

[a] $p<0.05$: significant difference in comparison to bipedic stance and stance with toe-extension.
[b] $p<0.05$: significant difference in comparison to Medial-Lateral Stability Index (MLSI) and Anterior-Posterior Stability Index (APSI) among the same postural condition.

Table 4.3: Descriptive statistics for the balance stability mean score for female.

Conditions	Female		
	Mean ± SD	95% CI	CV (%)
Biepdic Stance			
OSI (˚)	0.008 ± 0.003	0.006-0.009	32.3
MLSI (°)	0.003 ± 0.001 [c]	0.002-0.004	47.1
APSI (°)	0.006 ± 0.003 [b]	0.005-0.007	43.0
Stance with toe-extension			
OSI (˚)	0.008 ± 0.003	0.007-0.010	36.3
MLSI (°)	0.005 ± 0.002 [a]	0.003-0.006	55.6
APSI (°)	0.005 ± 0.002	0.004-0.007	35.9

[a] $p<0.05$: significant difference in comparison to bipedic stance and stance with toe-extension.
[b] $p<0.05$: significant difference in comparison to Medial-Lateral Stability Index (MLSI) and Anterior-Posterior Stability Index (APSI) among the same postural condition.
[c] $p <0.05$: significant difference between gender group among the same postural condition.

Table 4.4: Descriptive statistics for the balance stability mean score for all subjects.

Conditions	Overall		
	Mean ± SD	95% CI	CV (%)
Bipedic stance			
OSI (˚)	0.007 ± 0.002	0.006-0.008	30.8
MLSI (°)	0.001 ± 0.002	0.002-0.003	45.7
APSI (°)	0.006 ± 0.002 [b]	0.005-0.006	39.8
Stance with toe-extension			
OSI (˚)	0.008 ± 0.003	0.007-0.009	37.8
MLSI (°)	0.003 ± 0.005 [a]	0.004-0.006	56.0
APSI (°)	0.005 ± 0.002	0.004-0.006	37.1

[a] $p<0.05$: significant difference in comparison to bipedic stance and stance with toe-extension.
[b] $p<0.05$: significant difference in comparison to Medial-Lateral Stability Index (MLSI) and Anterior-Posterior Stability Index (APSI) among the same postural condition.

Females displayed a greater sway than males across all the conditions, although no significant differences were observed between the OSI, MLSI and APSI scores. When comparing the data of BLS and SWT conditions within the male group (n = 15), significant differences were found between the OSI with APSI, the OSI with MLSI, and the APSI with MLSI (main effect $p < 0.001$). Meanwhile, for the female group (n = 15), the results exhibited similar findings (main effect $p < 0.001$). No differences were observed between the BLS and SWT conditions between the male and female groups. Additionally, no participant grasped the holding rails to regain balance during the tests.

4.4 Discussion

There have been numerous studies that have investigated balance stability for various standing postures. However, to date, there is no research in existence that is focused on a bilateral support stance with toe-extension. In this current study, the differences in postural stability between BLS and SWT, and the impact of gender on postural control, were explored.

The results of this study substantiate the hypothesis and show that there was no difference in postural sway in BLS and SWT conditions in the OSI, MLSI and APSI. The balance performance as represented by the OSI, MLSI and APSI stability scores was slightly lower during BLS. No significant difference in stability scores was found between the conditions of BLS and SWT. Slight differences in the BLS and SWT scores may be due to the fact that SWT requires more sway from the CoP as the body regains balance, with a smaller BoS. According to the windlass mechanism, when the toes are extended to their limit, the plantar aponeurosis that wraps around the metatarsal bone tightens and therefore causes the arch to rise, further lifting the metatarsal head (Hicks, 1954). Hicks (1954) claimed that the metatarso-phalangeal (MTP) head will increase the pressure of the toe against the ground and the height of the rising arch is related to the distance of the metatarsal head to the calcaneus. The current study demonstrated that the arch of the foot did rise and that the sway in both ML and AP directions insignificantly increased during the test as the MTP head extended. Hence, this is inconsistent with the previous findings of Hicks (1954) which stated that toe-extension of the forefoot increases the postural sway due to the foot pressure that is applied at the MTP joints. As such, the toe extension may result in a reduction of the base of support and this subsequently deteriorates the stability of the body (Watkins, 1999).

In spite of being closely matched for age (p = 0.892), the current study showed that gender does not influence human balance in the BLS and SWT conditions, since no differences in the stability scores were found between the male and female participants. This is in contrast with previous studies, which reported that males had a greater postural sway than females during quiet standing (Mickle, Munro, & Steele, 2010). Lee and Lin (2007) suggested that the difference in body weight of males will result in greater CoP execution during the single-leg standing task. However, the findings of Mickle, Munro, and Steele (2010) claimed that males executed greater postural sway, even though the body weights for both genders are nearly similar.

According to the findings, the ML sway among females was slightly higher, but this was not significant when compared with the male control group during both the BLS and SWT conditions. Meanwhile, the current findings show that an increased complexity of standing posture leads to an increase in the ML sway. The ML sway was different to the AP sway during BLS. There was a significant increase in the ML sway that was required to achieve balance, as the ML sway was similar with the AP sway in SWT. This finding may relate to the Q-angle. Studies have shown that females have a greater Q-angle than males due to the length of their femur and their bigger pelvis area (Herrington & Nester, 2004; Omololu & Ogunlade, 2009). A larger Q-angle for females may result in an increase of rotation in hip movement, since the CoM needs to be maintained within the base of support to achieve body balance (Powers, 2010). As such, greater sway will be generated in the ML direction and sideways sway will increase in response to this in order to regain body equilibrium. Maki, Holliday, and Topper (1994) also demonstrated that an increase in body sway in the ML direction might lead to an increased risk of falling. Interestingly, the findings also revealed that the value of APSI was slightly higher than MLSI across all conditions, although it showed differences in BLS and no statistically significant difference in SWT. This is in agreement with previous studies which revealed that MLSI has a low value compared to APSI (Hermens et al., 1999; Ku et al., 2012b; Pereira et al., 2008). Meanwhile, the current findings show that there is an increase in the ML sway in accordance with the increasing complexity of the standing posture. The ML sway was found to be different to the AP sway during BLS. There was a significant increase in the ML sway that was required to achieve balance, as the ML sway was similar to the AP sway in SWT. As such, the tendency of the ML sway was considered to be an essential component for balance equilibrium when a more complex posture was applied during quiet standing.

The investigation of static balance control provides normative stability data for clinicians who are concerned with this specific toe condition. It is important to identify the changes of balance performance during SWT, since it is altered by the base of support and CoM. Some authors reported that the CoP and CoM are approximately equal only in the static or quasi-static conditions (Hasan et al., 1996). Likewise, Murray, Seireg, and Scholz (1967) presented the idea that motion in the CoP is greater than the CoM in order to keep the CoM within the base of support, since the CoP changes in response to the CoM. From the anatomical point of view, the height of the CoM is normally lower in females than males (Fessler, Haley, & Lal, 2005). There are also studies that investigate the effect of the type of female's footwear on balance (Cronin, Barrett, & Carty, 2012; Csapo, Maganaris, Seynnes, & Narici, 2010). High heeled shoes tend to affect postural control by raising and shifting the CoM forward (Menant, Steele, Menz, Munro, & Lord, 2008). Csapo et al. (2010) highlighted that the long-term use of high heels might diminish the Gastrocnemius muscle fascicles and reduce the range of motion in an individual's ankle, resulting in a feeling of discomfort even when they are wearing flat shoes.

Although the current study was restricted to an investigation of static postural control during toe-extension standing activities, it would be interesting to compare the postural control between active toe-extension in healthy individuals and the passive toe-extension that can be caused by burn injuries. Nevertheless, the investigation of static postural control has provided a normative stability data in SWT. Moreover, the sample size in this study only involved young adults (aged between 19 and 25 years old) from one specific geographical area and, as such, the data is limited.

The findings in this chapter have demonstrated thatthe differences between the BLS and SWT conditions do not lead to differences in postural control. However, a small alteration in postural control during SWT shows a trend of a greater amount of postural sway than BLS, although this is not significantly different. The findings also revealed the interactive effect of postural control in terms of the OSI, MLSI and APSI. Gender does not appear to effect static postural stability. The content in this chapter was reproduced from Ku et al. (2012b), with the authors' agreement. For Chapters 3 and 4, the baseline of balance performance in young adults was established. The following chapter will focus on the changes of postural stability and muscle activation in middle-aged adults during static and dynamic stance conditions.

CHAPTER 5:
THE LIMITS OF STABILITY AND MUSCLE ACTIVITY IN MIDDLE-AGED ADULTS DURING DYNAMIC STANCE

5.1 Introduction

Middle-age is the life stage transition between young adult and elderly, which is experiencing changes of physical, biological and psychologically (Jeong, 2010). Aging process naturally deteriorates sensory systems, muscle mass, and further impacts general balance ability for most of the middle-aged adults. Poor balance ability is related with the risk of falling and lower extremity injury (Maki et al., 2011; Winter, 1995). Falls may cause bone fractures, head injuries, induce fear of falling, and further impact on the quality of life (Alexander, Rivara, & Wolf, 1992; Sterling, O'Connor, & Bonadies, 2001).

The changes of balance ability in middle-age adults may be associated with the deterioration of the musculoskeletal system, cognitive tasks, visual and vestibular systems (Seidler et al., 2010). The process of aging in middle-aged deteriorates many physiological systems (Błaszczyk & Michalski, 2006), with reaction time, speed of learning, and functional mobility being significantly lower than in younger adults (Bernard-Demanze, Dumitrescu, Jimeno, Borel, & Lacour, 2009). Fransson, Kristinsdottir, Hafstrom, Magnusson, and Johansson (2004) reported that middle-aged adult tends to have better responses latency and motion complexity with eyes open than the elderly. Moreover, Hunter, Thompson, and Adams (2000) indicated that higher physical activity delays the impairment of age-related changes in muscle strength, and the risk of obesity and falling would be reduced. Anderson, Deluigi, Belli, Tentoni, and Gaetz (2016) reported increased of core muscle activation after short term training program, suggesting an improvement of muscle coordination and efficiency among middle-aged women.

Static and dynamic stance balance are important elements used to assess the independence of daily living activities of an individual. Dynamic stance balance is crucial for the acquisition and execution of motor control, and depends on the control of CoG, weight shifting, and active muscle control (Melzer, Benjuya, Kaplanski, & Alexander, 2009; Pugh, Heitman, Kovaleski, Keshock, & Bradford, 2011). Limits of stability (LoS) is a reliable tool for dynamic stance control for fall risk screening and functional stability amongst the healthy adult population (Juras, Fredyk, Sobata, & Bacik, 2008). The stance control of LoS refers to the ability of the maximum voluntary distance or angle in which an individual can regulate the CoG in a given direction, without losing balance (Horak, Dimitrova, & Nutt, 2005). It is considered as a boundary involving functional, anatomic, mechanical limit, and internal body shifting for maintaining stability without changing the base of support (Kolarova et al., 2013).

To date, many studies have investigated the upright stance performance amongst the general population, mostly focusing on young adults and the elderly. Only very limited research has analysed middle-aged adults. Such adults are an important life stage, as they face the problem of balance ability declination. Thus, the study aim in this chapter was to determine the changes of directional stability and lower limb muscle activation in static and dynamic stance, for sedentary middle-aged adults. The symmetry of directional stability and muscle activation between left and right legs was also examined. It is hypothesized that directional stability would decrease for a greater tilt angle of moving surface, and that muscle activation and displacement of CoP would be symmetrical between left and right legs.

5.2 Methods

5.2.1 Participants

Twenty middle-aged adults (10 males, 10 females) were recruited for this study (mean ± standard deviation, age= 50.0 ± 7.5 years; body height: 1.61 ± 0.10 m; body weight: 70.0 ± 14.5 kg) (Table 5.1). The inclusion criteria for the participants were (i) aged over 40 to 60 years, (ii) community active and independent walkers with sedentary lifestyle, (iii) with normal vision, (iv) able to perform normal activities independently, (v) had no history of neurological, visual, vestibular or balance impairments (vi) had no history of surgery to the lower limb, and (vii) with right dominant leg. The study was approved by the University of Malaya Medical Ethics Committee (MEC 895.7). After providing an explanation about experimental procedures and risks to the subjects, they gave written consent for this study. Before the experiment commenced, participants were required to complete the Berg Balance scale to evaluate their daily balance functionality. The average Berg Balance score obtained for this study was 55.9 ± 0.3.

Table 5.1: Participant anthropometric characteristics.

Variables	Men (*n*=10)	Women (*n*=10)
Age, years	48.4 ± 12.1	50.1 ± 4.2
Weight, kg	74.1 ± 10.9	65.9 ± 74.1
Height, m	1.68 ± 0.05	1.54 ± 0.08
BMI, kg/m^2	26.2 ± 3.6	27.8 ± 6.7
Berg Balance Score	56 ± 0	55.9 ± 0.32

5.2.2 Instrumentations

BBS was used to measure the displacement of CoP. It is composed of a circular platform which allows 0-20˚ of platform surface tilt in a 360˚ range of motion. Data were sampled at a frequency of 100 Hz. A microprocessor-based actuator was used to adjust the level of resistance of 8 springs, in order to control the tilt degree of the circular platform for stability limit. The spring is made from music wire with a spring rate of 13.81 N/cm. The spring compression generates 88.9 N of force (Arnold & Schmitz, 1998a). For the LoS test, there are 12 different dynamic stability levels for the circular platform: from levels 1 (most unstable) to 12 (most stable). The measured parameter for LoS was directional control (DCL) scores for 8 predetermined target positions in the forward, backward, rightward, leftward, forward-rightward, forward-leftward, backward-rightward, and backward-leftward directions. Figure 5.2 shows the calculation method for straight line distance to target for LoS. The formulas for LoS are shown as Equation 5.1.

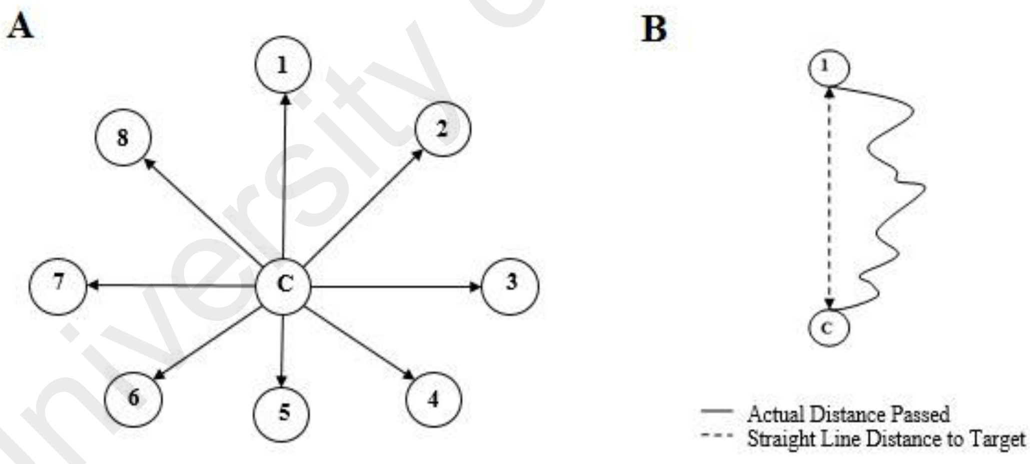

Figure 5.1: The protocol of the LoS test. (A) LoS directions are defined as 1: forward; 2: forward-rightward; 3: rightward; 4: backward-rightward; 5: backward; 6: backward-leftward; 7: leftward; 8: forward-leftward. (B) The calculation of LoS with the resultant CoP displacement in the BBS.

$$Score\ \% = \frac{Straight\ Line\ Distance\ to\ Target}{number\ of\ samples}\ x\ 100 \qquad 5.1$$

Muscle activities were analyzed with the Trigno™ Wireless EMG System (Trigno™ Wireless, Delsys Inc., Boston, MA), using Trigno parallel-bar surface sensors (37x26x15 mm^3, Trigno™ Wireless, Delsys Inc.). After the skin was cleaned and swabbed with alcohol to minimize skin resistance, electrodes with an adhesive skin interface were placed in the appropriate position. According to the Surface Electromyography for Non-Invasive Assessment of Muscles Recommendations, electrodes were placed over left and right rectus femoris, biceps femoris, and medial gastrocnemius muscle groups.

5.2.3 Study Protocol

All tasks for LoS in the BBS were randomized to eliminate order effects. Participants stood barefoot on the Biodex Balance System with attached electrodes, with hands placed by their side. The foot position was standardized in order to minimize measurement error. The participant's footprints were marked for the consistency of consecutive trials. Once the test commenced, participants were instructed to lean as far as possible in a particular direction, with real-time feedback illustrated on the Biodex Balance System display screen. Eight different directions were used for the LoS. Participants were required to control the cursor (shown on the display screen) to coincide with the target, without lifting their feet or flexing their hips. A minimum of 2 s holding time in the leaning position was required before participants could return the cursor back to the initial position at the center. The stability level of LoS is defined as the stiffness of the foot platform (Biodex Medical Systems, 1999). Increased stability level represents the increase in the tilt angle of the moving platform. Platform stiffness decreases as stability level increases. The experimental task was performed using five different platform stability levels: 4, 6, 8, 10 and 12. This allowed 14º, 11º, 7º, 4º and 0º of deflection from a level in any direction. Level 12 was set as the control level for this study. Two practice trials for each level were performed before the test commenced. Three consecutive trials were performed under each level of LoS, and a 1 min rest period was allowed between trials. All participants were fitted with a harness system to prevent falls during the test.

5.2.4 Data Analysis

For LoS, DCL scores for rightward, forward-right and backward-rightward were averaged as the right side of DCL. Scores for leftward, forward-leftward and backward-leftward were averaged as the left side of DCL. The right and left side of DCL were used to determine the left-right asymmetry of DCL for different stability levels.

For all electromyography (EMG) channels, the raw EMG signal was collected at 4000 Hz and stored using EMG work® Acquisition software (Version 4.1.7, Delsys Inc., Boston, MA). The raw EMG signal was filtered using a 4th order Butterworth filter with 20-450 Hz filter to eliminate movement artefacts. For normalization, the root mean square (RMS) for 30 s resting-state muscle activation in sitting posture was obtained as the baseline value. After measured EMG data from the first and last seconds were excluded, the measured EMG values at the dynamic-stance state was converted to RMS. All RMS values calculated for rectus femoris, biceps femoris and medial gastrocnemius at the dynamic-stance state were normalized with baseline values, in order to provide an expression of relative muscle activation. The mean RMS EMG values of three consecutive trials was averaged. A comparison of muscle activation between left and right leg was computed for rectus femoris, biceps femoris and medial gastrocnemius.

5.2.5 Statistical Analysis

Data were expressed as mean values ± standard deviations. The Kolmogorov-Smirnov test was used to identify data normality of distribution. The analysis of normality test showed that all data was normally distributed. Independent sample t-tests were conducted to identify differences due to gender, and differences between left and right leg of DCL and muscle activity. The alpha level for the *t*-test was set to 0.05. Two-way repeated measures analysis of variance (ANOVA) with 5 x 8 was used to examine differences between stability level and DCL. In addition, the two-way ANOVA with 5 x 6 (stability level x muscle activation) was tested to evaluate any significant effects associated with stability levels and lower limb muscle activation. The *F*-ratio correction was applied with Greenhouse-Geisser (Epsilon, $\varepsilon < 0.75$) and Huynh-Feldt tests ($\varepsilon > 0.75$) if the data violated the sphericity assumption ($p < 0.05$) in Mauchley's test. The *post hoc* Bonferonni test was implemented to identify significant differences for multiple comparisons. The alpha level for the ANOVA was set at $p = 0.002$ after Bonferroni correction. Statistical tests were executed by using SPSS version 22.0 (SPSS for Windows, IBM Inc., Chicago, IL, USA).

5.3 Results

5.3.1 Direction Control in LoS

No significant differences between genders were observed for age, body mass, and body mass index. Table 5.2 provides the mean and standard deviation of DCL for different moving platform tilt angles. Accordingly, the ANOVA demonstrated a significant main effect of directional stability ($F_{7,\ 133} = 2.75$, $p = 0.011$) and main effect of stability level ($F_{2.62,\ 49.73} = 11.64$, $p < 0.001$). The DCL for level 4 was significantly lower compared to level 6 ($p = 0.004$), level 8 ($p = 0.012$), and level 12 ($p = 0.001$). For the DCL difference between control level and level 4, the LoS score of level 4 was significantly lower in rightward (p = 0.001), leftward (p < 0.001), and backward-rightward directions (p = 0.001). There were no interaction effects between directional stability and stability level. Further, no left-right asymmetry of DCL nor gender effect for directional stability was found in all five stability levels (Table 5.2). Moreover, the task-completed times for subjects were longer (p < 0.001), whilst the overall LoS score was decreased significantly (p < 0.001) when the platform tilt angle increased.

5.3.2 EMG activity

Figure 5.2 – 5.4 presents the mean RMS of EMG activity for all muscle groups during five different stability levels. The ANOVAs indicated a significant main effect of muscle activity for different muscle groups ($F_{2.27, 43.20} = 4.47$, p = 0.014). Based on the multiple-pairwise comparison between the muscle groups, a significant difference was found between the left-right biceps femoris (p = 0.003), right biceps femoris –right rectus femoris (p < 0.001), and right biceps femoris –left rectus femoris (p = 0.022). No main effect of stability level, and interaction effect between stability level and lower limb muscle activity were observed in the study. For the comparison of muscle activation within the lower limb, the left biceps femoris was significantly greater than the right biceps femoris for all stability levels (Figure 5.3). The right medial gastrocnemius activity was significantly greater at stability levels 8, 10 and 12 (Figure 5.4). No gender effects with respect to lower limb muscle activation were found (Table 5.3).

Table 5.2: Directional control scores for limits of stability test in different dynamic stability levels.

	Dynamic Stability Level (°)									
	L12	Range	L10	Range	L8	Range	L6	Range	L4	Range
Complete time (s)	47.1(6.0)	37.3-59.0	58.4(8.9)	43.3-74.3	60.3(8.0)	49.3-75.7	60.1(11.1)	37.7-80.7	64.7(17.8)	33.5-103.7
Overall (%)	51.0(11.7)	29.0-72.7	42.0(9.8)	26.0-59.0	45.1(11.7)	29.0-67.3	41.2(12.7)	19.0-69.3	32.2(16.5)	11.7-63.7
Forward (%)	54.3(16.8)	19.0-77.7	49.9(13.2)	30.7-81.0	50.6(17.2)	19.3-85.0	50.0(14.1)	22.0-74.7	41.3(21.2)	9.7-80.7
Backward (%)	57.3(20.6)	28.3-97.3	48.5(14.1)	18.3-68.3	57.0(18.1)	33.0-93.0	54.3(20.8)	25.7-90.0	39.1(22.3)	7.3-80.0
Right (%)	61.7(15.6) *,†	32.3-92.0	53.9(15.0)	22.3-77.3	52.1(12.8)	34.0-78.3	42.4(11.8)	19.7-64.0	41.7(20.7)	8.7-77.7
Left (%)	51.1(13.8)*	34.3-87.3	47.6(12.3)	16.7-69.0	52.4(13.8)	23.3-77.0	46.9(19.1)	15.7-79.7	37.3(15.1)	11.0-62.7
Forward/Right (%)	53.5(14.9)	28.7-77.7	47.1(13.8)	24.3-79.0	49.3(12.5)	26.0-69.3	47.1(17.2)	19.7-75.0	40.1(15.8)	14.0-76.7
Forward/Left (%)	59.2(14.5)	30.0-80.7	50.8(15.5)	26.0-79.7	52.3(13.7)	25.7-75.3	49.6(15.5)	24.3-76.7	40.3(21.4)	6.7-80.0
Backward/Right (%)	54.9(17.3)*	15.0-86.0	42.1(13.5)	22.3-77.7	48.0(12.0)	27.3-69.7	45.7(14.1)	23.3-78.7	35.7(18.4)	10.3-69.0
Backward/Left (%)	52.1(16.0)	19.3-86.7	45.6(11.8)	27.3-68.3	48.4(19.5)	19.7-84.0	51.4(17.0)	22.0-83.7	40.8(18.2)	18.3-67.3

* $p<0.002$, indicates significant difference of directional stability compared to level 4.
† $p<0.002$, indicates significant difference of directional stability compared to level 6.

Table 5.3: The directional control scores and muscle activity for male and female in different dynamic stability levels.

	Level 12			Level 10			Level 8			Level 6			Level 4		
	Male	Female	Cohen's *d*	Male	Female	Cohen's *d*	Male	Female	Cohen's *d*	Male	Female	Cohen's *d*	Male	Female	Cohen's *d*
Directional Score															
Left *	58.3(15.6)	48.4(18.3)	0.59	47.3(13.3)	56.0(16.2)	0.62	47.5(16.1)	54.5(14.8)	0.45	42.5(15.4)	48.8(13.4)	0.43	30.5(13.1)	56.6(14.7)	1.87
Right †	56.9(15.0)	48.3(17.4)	0.53	46.3(12.2)	48.9(14.6)	0.19	48.3(11.7)	51.3(13.0)	0.24	41.3(13.4)	49.0(16.9)	0.50	30.0(14.2)	56.4(17.4)	1.66
Biceps Femoris															
Left	1.41(0.55)	2.14(1.90)	0.52	1.69(0.51)	2.11(1.08)	0.50	1.71(0.61)	1.94(1.11)	0.26	1.76(0.65)	1.85(1.06)	0.10	2.13(1.14)	2.17(1.70)	0.03
Right	1.00(0.03)	1.03(0.04)	0.85	1.18(0.51)	1.03(0.06)	0.41	1.01(0.02)	1.03(0.06)	0.45	1.01(0.02)	1.03(0.01)	1.26	1.02(0.03)	1.15(0.38)	0.48
Rectus Femoris															
Left	1.21(0.42)	1.45(0.63)	0.40	1.73(1.00)	1.89(1.39)	0.13	1.75(1.00)	1.66(0.73)	0.10	1.80(1.04)	1.70(0.69)	0.11	2.26(1.42)	2.09(0.94)	0.14
Right	1.37(0.29)	1.26(0.28)	0.39	1.66(0.50)	1.40(0.18)	0.69	1.65(0.49)	1.40(0.23)	0.65	1.64(0.42)	1.44(0.25)	0.58	1.75(0.53)	1.66(0.37)	0.20
Gastrocnemius															
Left	1.23(0.29)	1.24(0.34)	0.03	1.29(0.37)	1.24(0.35)	0.14	1.27(0.37)	1.28(0.62)	0.02	1.25(0.38)	1.49(0.84)	0.37	1.31(0.42)	1.25(0.41)	0.14
Right	2.42(2.11)	2.20(1.43)	0.12	2.68(1.71)	2.03(1.18)	0.44	2.37(1.56)	1.94(1.13)	0.32	2.17(1.84)	1.87(0.98)	0.20	1.77(1.11)	1.74(0.85)	0.03

* The sum of directional score for leftward, forward-leftward, and backward-leftward.

† The sum of directional score for rightward, forward-rightward, and backward-rightward

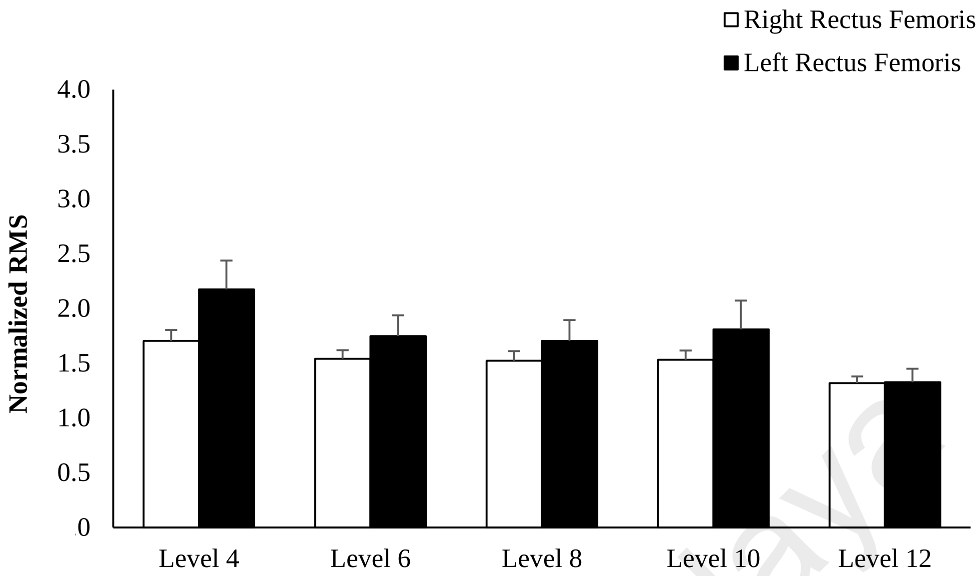

Figure 5.2: The mean RMS of EMG activity in middle-aged adult during five different dynamic stability level for rectus femoris muscle group.

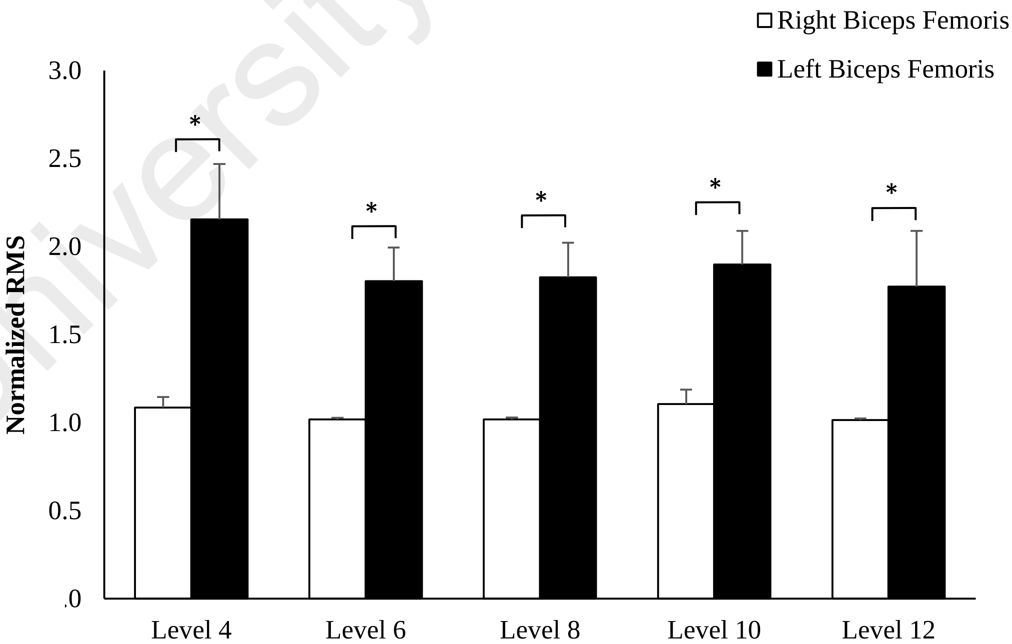

Figure 5.3: The mean RMS of EMG activity in middle-aged adult during five different dynamic stability level for biceps femoris muscle group.* $p < 0.002$

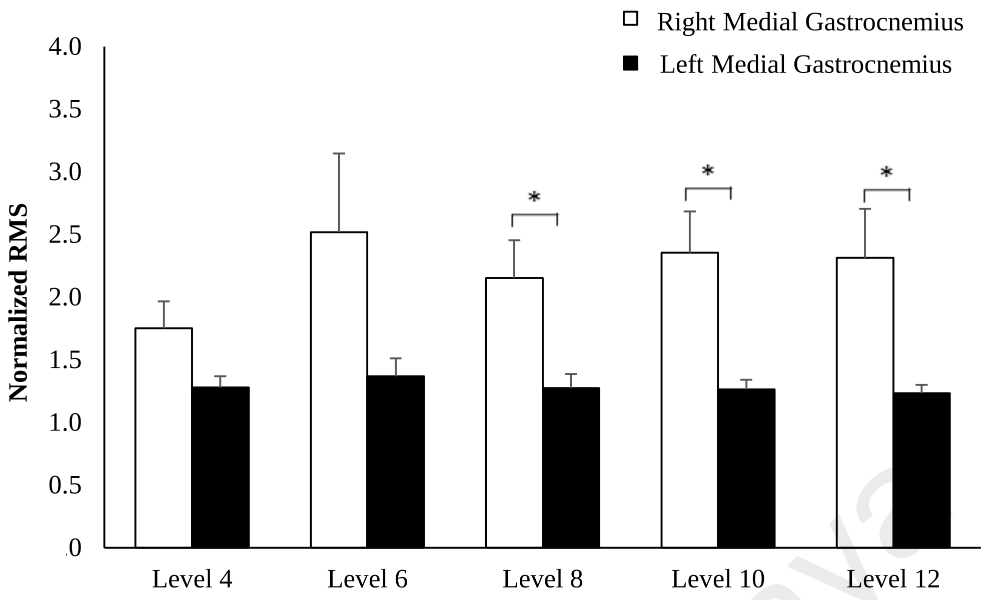

Figure 5.4: The mean RMS of EMG activity in middle-aged adult during five different dynamic stability level for medial gastrocnemius muscle group. .* $p < 0.002$

5.4 Discussion

The purpose of this study was to examine changes of directional stability and lower limb muscle activation with a different tilt surface amongst middle-aged adults. The symmetry of directional stability and muscle activation between left and right leg was also investigated. The present findings revealed the significant difference of DCL for the increased tilt angle for the dynamic platform. Directional stability was significantly declined as the tilt angle of the moving platform increased from 0º to 14º, with exception of forward, forward-rightward and backward motions. This study also demonstrated that the right biceps femoris activation was significantly lower than left biceps femoris, left rectus femoris and right rectus femoris. Muscle activation significantly increased as the platform tilt angle increased for right rectus femoris, and was practically significant for left rectus femoris. In addition, the muscle activity of left-right asymmetry was found significantly different in biceps femoris and medial gastrocnemius muscle groups.

As expected, the increase in stiffness of the platform significantly increased the CoP sway in particular directions. This is in line with the studies of Pugh et al. (2011) and Perron, Hebert, McFadyen, Belzile, and Regniere (2007), with the tilt angle of a moving platform increasing balance difficulties. By comparing the current findings with similar study, healthy young adult tends to have a lower DCL score compared to middle-aged adult in level 4, 6 and 8 (Perron et al., 2007). Faraldo-Garcia, Santos-Perez, Crujeiras, and Soto-Varela (2016) revealed that the DCL scores for forward, backward, leftward and rightward were gradually reduced as the age group increased. Age effect have deteriorates the muscle strength and functional control on the task performing, and impairs the balance performance. Moreover, the body leans outward from the original upright position inducing CoG motions in both the anteroposterior and mediolateral axis. In this regard, the current findings demonstrate changes of dynamic directional stability in leftward, rightward and backward leaning when the platform stiffens, whilst forward DCL remains constant. It is speculate that with the limited range of motion at the ankle joints, increased platform stiffness does not have significant impact on forward DCL. For dynamic stance control, anteroposterior movement is regulated by the ankle joint, with mediolateral movement being controlled by the hip joint (Horak & Nashner, 1986; Nashner, 1976). Postural sway during forward trunk leaning was affected by ankle plantar flexors. The ankle joint generates maximal plantarflexion and dorsiflexion torques for reducing the body’s forward and backward leaning instabilities, respectively. Within the range of motion for the ankle (0º-50º), adequate force and motion can easily allow ankle plantarflexion to exert a downward movement and lean the body forward in the dynamic platform. Further, changes of weight loading were related to the directional leaning. Kanakis, Hatzitaki, Patikas, and Amiridis (2014) reported that leaning direction was affected by the weight loading of ankle muscles. At the ankle, the weight distribution that loads on medial gastrocnemius tends to exert a higher activation of plantarflexor, compared to backward leaning during upright stance. Forward trunk leaning from the upright stance may increase spinal loading (Takahashi, Kikuchi, Sato, & Sato, 2006). Forward leaning with arm support assists ribcage elevation and optimizes the recruitment of accessory muscle (Gosselink, 2003).

The present findings indicate that there is a suggestive trend that rectus femoris activity increases significantly with the increase of dynamic platform tilt. Rectus femoris is the main quadriceps muscle regulating the contraction for hip flexion, and the biceps femoris muscle is the antagonist. It is thought that increased rectus femoris activity is related to the upright position. The extended knee joint due to upright body alignment will lock the knee joint, whereas the hip joints act to compensate the inadequate torque of the ankle joint to lean the body laterally in the dynamic stance. The reduction of rectus femoris activity will be induced by inclining the body on the desk level in upright position (Damecour, Abdoli-Eramaki, Ghasempoor, & Neumann, 2010). Besides rectus femoris and biceps femoris, de Freitas, Freitas, Duarte, Latash, and Zatsiorsky (2009) analyzed the contribution of the hip abductor in lateral stability, where the gluteus minimus and maximus control pelvic tilting in the anteroposterior axis, with the gluteus medius controlling pelvic tilting in the mediolateral axis.

The deterioration of sensory and motor skills with age impact stability limits and muscle activity (Melzer et al., 2009; Teasdale & Simoneau, 2001). Melzer et al. (2009) found that the reduction of ankle muscle strength diminishes LoS capacity amongst the elderly. Laughton et al. (2003) revealed that, with their increased anteroposterior sway and muscle activation, the elderly tend to fall more easily compared to young adults.

Interestingly, the present study found that individuals with a right dominant leg tend to increase their weight-shifting in forward-rightward and backward-leftward leaning. Additionally, the left-right asymmetry was only found in biceps femoris and medial gastrocnemius muscles. Previous studies have indicated no impact of leg dominance on postural sway for the stance posture (Harrison, Duenkel, Dunlop, & Russell, 1994). However, the significance of left-right asymmetry could be explained by the greater muscle strength of the dominant leg, compared to non-dominant (Jacobs, Uhl, Seeley, Sterling, & Goodrich, 2005). In the upright stance mechanism, the medial gastrocnemius muscle is predominantly used for forward body leaning, whereas biceps femoris muscle is used for backward leaning. The immediate change of CoG for the dynamic voluntary inclination may simultaneously activate calf and biceps femoris muscles. Huurnink, Fransz, Kingma, Hupperets, and van Dieen (2014) suggested that that the dominant leg could not be standardized for all types of task, as it will change according to the task applied.

There are limitations in this study. Firstly, the sample size is relatively small, and limited to middle-aged adults with a sedentary lifestyle. Improved fitness level would augment maximum overall muscle activation and strength (Hakkinen et al., 1998). Secondly, the findings may have been affected by the gender group. Even though no difference between genders was found in this study, body shape and morphological differences may have potentially impacted the results. Thirdly, fear of falling could impact dynamic balance performance. Although a harness system was fitted on participants during the test, they may still have been fearful of falling down.

This study provided the changes of postural control for upright stance patterns and muscle activity in middle-aged adult for the improvement of independent dynamic control, in order to allow them to actively prepare for their elderly life ahead. In conclusion, middle-aged adults exhibited a lower stability limit for directional balance control as the stiffness of the dynamic platform increased. In addition, only the rectus femoris muscle exerted a significant increase in muscle activation with the increase of dynamic stability level. The activity of biceps femoris and medial gastrocnemius muscle appears to exert a left-right asymmetry for the lower limb muscle, while no left-right asymmetry of directional stability was observed. Whilst the present findings reflect the postural control of such adults, further studies are required to investigate the postural control and muscle activation for age-matched patient groups with balance abnormalities. The following chapter will investigate more on the balance performance on middle-aged transtibial amputees. In addition, the effect of muscle activity for static and dynamic stance control also will be examined.

CHAPTER 6:
THE DIRECTIONAL STABILITY AND LOWER LIMB MUSCLE ACTIVITY DURING DYNAMIC STANDING AMONG TRANSTIBIAL AMPUTEES

6.1 Introduction

Transtibial amputation is defined as the removal part of the lower limb below knee due to trauma, peripheral vascular disease, tumor, infection or other causes (Zidarov, Swaine, & Gauthier-Gagnon, 2009). Following the transtibial amputation, the missing ankle joint leads to the development of a new control mechanism that compensates for the habitual stance mechanism (Vrieling et al., 2008). Previous studies have found that load distribution of the lower limb amputee is likely to veer towards the side of the intact leg (Nadollek, Brauer, & Isles, 2002; Nederhand et al., 2012), and the CoP excursion for the intact leg was also greater than the amputated leg during static stance (Duclos et al., 2007; Quai, Brauer, & Nitz, 2005). Kolarova et al. (2013) found that TTAs have a decreased voluntary body inclination backward within the stability limits. The restricted CoP displacement, inadequate directional control and reduced velocity will limit the control of CoG within the stability limits (Molero-Sánchez et al., 2015). Bolger, Ting, and Sawers (2014) suggested that the deficit in balance control for TTAs was related to the inability to control the prosthetic leg to exert adequate forces which are equal in magnitude and direction to healthy adults.

Although many studies investigated the postural balance of the lower limb amputee, few studies have focused on stance balance during a dynamic stance as the dynamic platform is a technique to study postural balance perturbation (Horak & Nashner, 1986). Therefore, the purpose of this study was to investigate the effects of dynamic stance on directional stability and lower limb muscle activity. It is hypothesized that platform stiffness would deteriorate the postural stance performance and a significant difference of muscle activities between the dynamic stance levels would be observed.

6.2 Methods

6.2.1 Subjects

Sixteen unilateral TTAs were recruited for the study. The subjects were included if they were: (i) aged 40-65 years, (ii) able to perform independent movement without walking aids for at least 30 min, (iii) experienced lower limb amputation and used a prosthetic limb on a daily basis for at least 1 year, (iv) no falling history within 1 year, and (v) no previous history of neurological, vestibular and visual impairments. The reason for amputation for all subjects in this study was traumatic (n=8) and peripheral vascular disease (n=8). Nine subjects had undergone right limb amputation, whilst seven subjects undergone left limb amputation. The subjects wore their own prosthetic limb for the experiment tasks in the study. To obtain the functional information, subjects were required to perform an Amputee Mobility Predictor with Prosthesis (AMPPRO) for the ability of locomotion in amputee. The average AMPPRO score is 29 ± 9.86. This study was conducted at the university performance laboratory. All subjects have provided written consent to participate in this study and ethical approval was obtained from the University Medical Committee (Ethic no: MEC 895.7). Table 6.1 shows the summary of TTAs' demographic characteristics in the present study.

Table 6.1: The participants' characteristics for TTA.

	Male	Female
No. subjects	10	6
Age (years)	48.6 ± 7.5	50.7 ± 4.3
Height (m)	1.70 ± 0.05	1.60 ± 0.05
Weight (kg)	73.81 ±15.98	66.33 ± 18.12
BMI (kg/m^2)	26.96 ± 4.71	25.81 ± 6.45
Time since amputation (years)	5.79 ± 3.38	8.50 ± 8.67

6.2.2 Procedure

Before the test commenced, surface electrodes were attached to relevant positions on the lower limbs of the subjects. All LoS for different stability levels were randomized to eliminate the order effects. The LoS tests were acquired with stability level 4, 6, 8, 10 and 12. Level 12 was set as control level since the moving platform was static. During LoS testing, subjects were stood on the circular platform with the hands placed alongside. The footprints on the platform were marked and recorded for consistency in repeated trials. Subjects were prompted to move the real-time feedback cursor when instructed from the liquid crystal display. Eight predetermined targets were set for different directions and randomly highlighted during the test. By leaning the body towards the target direction, it allows the cursor to coincide with the highlighted target. After reaching the target with a minimum holding time of 2 s, subjects were required to return their body to the center position. The dynamic LoS test was completed if all 8 predetermined target had been coincided. Three repeated trials were performed for each stability level. A resting time of 1 min between trials was allowed. Subjects were instructed to complete the test as quickly as possible. All subjects were prohibited to lift their feet or flex their hips during the test. Two practice trials for each stability level were allowed before the test commenced. For safety purposes, a harness system was fitted on subjects to prevent falling.

6.2.3 Limits of Stability

All postural balance measurement were performed on the BBS (Biodex Medical System Inc., Shirley, NY). The LoS test that provided in the BBS evaluates the directional stability with a circular platform which allows maximum of 20º surface tilt in 360º range of motion. In order to adjust the surface tilt, a microprocessor-based actuator located beneath the circular platform is used to control the level of resistance for 8 springs. The sampling rate is 100 Hz and the spring rate for the system is 13.81 N/m. The spring is made from music wire and will generate 88.9 N of forces after compression (Arnold & Schmitz, 1998a). There are 12 stability levels available for testing: level 1 (most movable) to level 12 (most stable). The platform stiffness will decrease as the stability level increases. The main measured parameter for 8 predetermined targets in LoS are complete time (s) and directional control scores (%). The results of 3 repeated trials were averaged in the analysis for each subject. The directional control score was expressed as a percentage score (Equation 5.1). A higher directional stability score represents better balance performance. The Intraclass Correlation Coefficient for Biodex measure of directional control defines as 0.72 and the standard error of the measurement values is ± 6.38% (Pickerill & Harter, 2011).

6.2.4 EMG Recording and Data Analysis

EMG signals were collected using parallel-bar sensor (37 x 26 x 15 mm^3, Trigno™ Wireless EMG System, Delsys Inc., Boston, MA) which placed at the skin of lower limb. The electrode's skin contact is fabricated with pure silver. Before the electrode placement, the skin was cleaned and an alcohol swab applied in order to minimize skin resistance.

The electrodes with adhesive skin interface were then placed over the left and right side of the rectus femoris, biceps femoris and medial gastrocnemius of the involved limb followed the surface electromyography for non-invasive assessment of muscle (Hermens et al., 1999).

EMG work® Acquisition software (Version 4.1.7) was utilized to store the raw EMG signal that was collected at 2000 Hz. All raw data was then analyzed using MATLAB software. To remove the movement artefacts, 4th order Butterworth with 20 – 450 Hz was used. A 30 s of sitting state muscle activation was recorded for data normalization. It was defined as the baseline value for the minimum muscle activation in the relaxed condition. The RMS of EMGs were calculated after the first and last second of EMG signal were eliminated. The RMS of EMG was normalized with the RMS of the baseline value and the RMS values of 3 repeated trials were averaged. Normalization aims to remove the minimum muscle activation in the stance position. The mean value of 3 trials was averaged for data analysis.

6.2.5 Statistical Analysis

All statistical procedures were performed with SPSS software (Window version 21; IBM Inc., Chicago, IL). The data normality that was screened by Shapiro-Wilks test was normally distributed in all analysis. The independent sample *t*-test was used to identify the difference of muscle activity between the left and right leg. To evaluate the mean differences of directional control, a 5 x 8 (stability level x direction) repeated measures ANOVA was performed. ANOVA with 5 x 6 (stability level x muscle group) was applied to investigate significant differences of muscle activation between stability levels. If the data violated the sphericity assumption ($p < 0.05$) in the Mauchley's test, the F-ratio correction was applied by the Greenhouse-Geisser test (Epsilon, $\varepsilon < 0.75$) and Huynh-

Feldt test (ε > 0.75). The Bonferroni *post hoc* comparisons were computed to locate the difference and correct for multiple comparisons when the significant *F*-ratio was found. Statistical significance was set at $p < 0.002$ after Bonferroni correction in ANOVA. Effect sizes (*d*) for each condition were calculated as the ratio of mean difference divided by its standard deviation. As effect size that was smaller than 0.2 was considered to be a small effect, 0.5 considered to be a moderate effect, and greater than 0.8 considered a large effect (Cohen, 1988). All data are reported as mean ± SD.

6.3 Results

6.3.1 Limits of Stability

ANOVA revealed a significant main effect of stability level ($F_{2.34,35.09} = 6.42$, $p = 0.003$). The directional control of stability level 4 has a significantly lower score than level 6 ($p = 0.023$), level 8 ($p < 0.001$), and level 10 ($p = 0.001$). Statistical analysis found a significant main effect between directions ($F_{7,105} = 6.30$, $p < 0.001$). No significant differences of LoS sway were found between the forward-backward and leftward-rightward directions. Only the forward direction showed a significantly smaller postural sway than the rightward ($p < 0.001$), leftward ($p < 0.001$), and forward-rightward directions ($p = 0.01$). In addition, the interaction effect between stability level and direction was statistically significant ($F_{28,420} = 2.06$, $p = 0.001$). The mean score of LoS for different directions with statistically significant findings is illustrated in Figure 6.1-6.4.

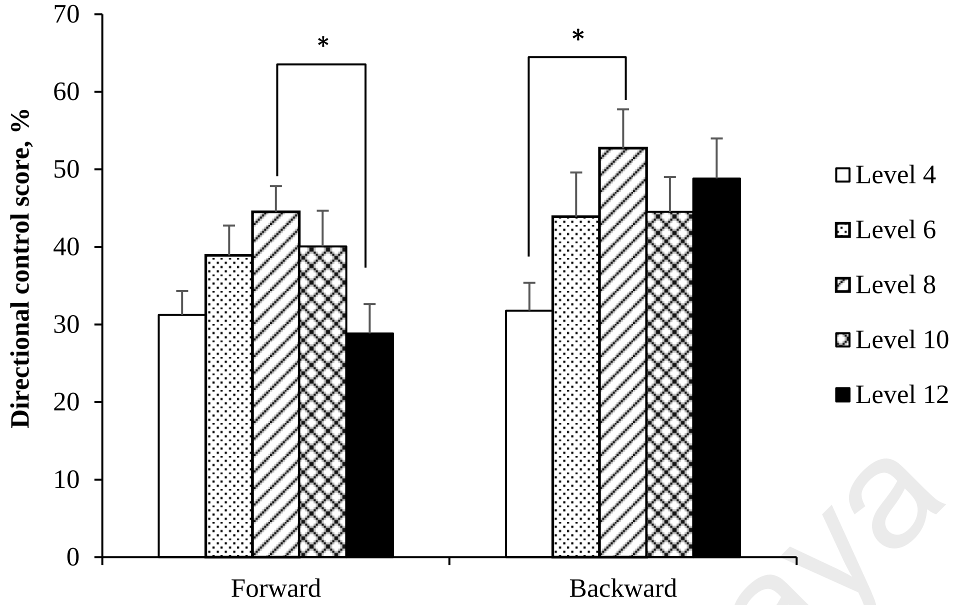

Figure 6.1: The directional control score of eight predetermined target direction for different stability levels in forward and backward direction, * p<0.002.

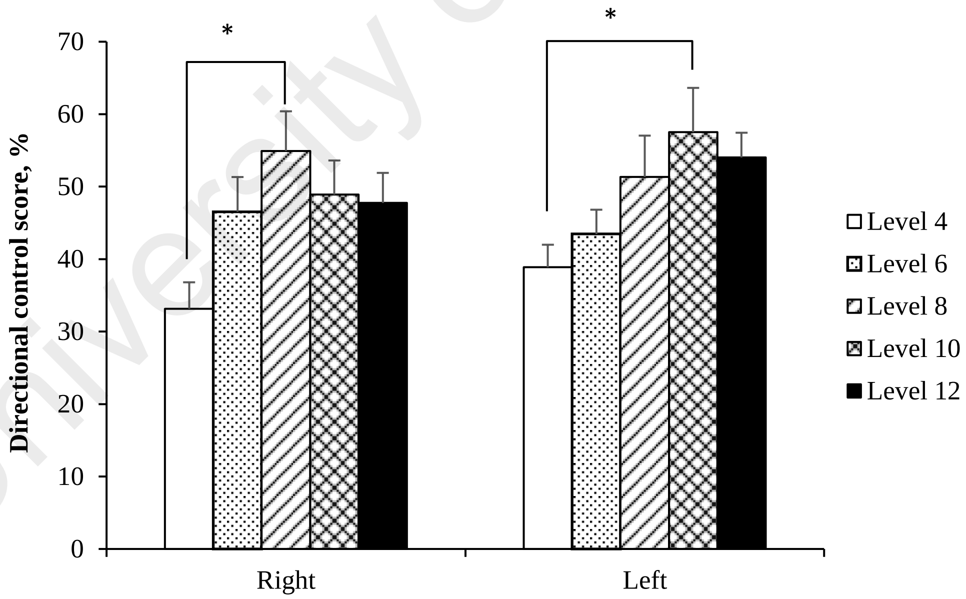

Figure 6.2: The directional control score of eight predetermined target direction for different stability levels in rightward and leftward, * p<0.002.

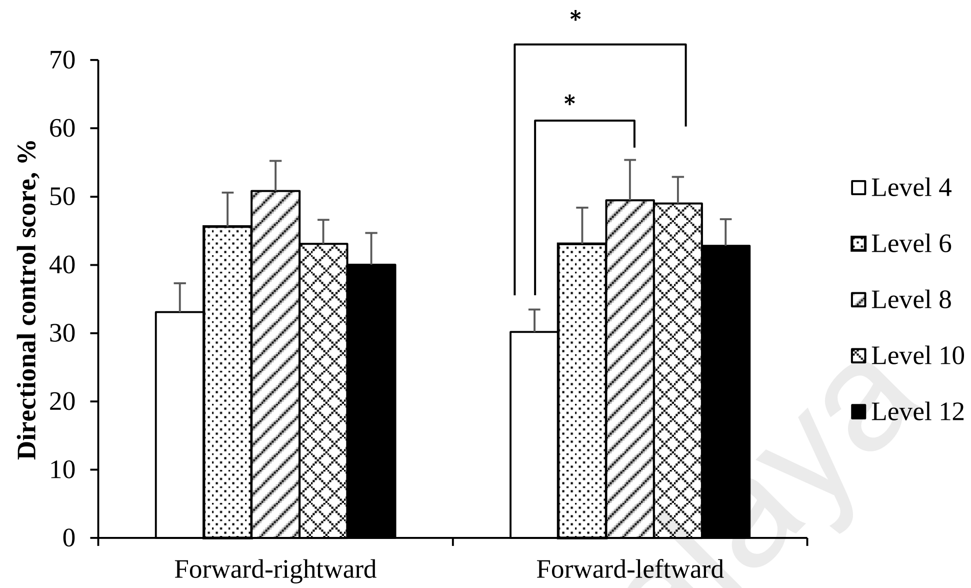

Figure 6.3: The directional control score of eight predetermined target direction for different dynamic stability levels in forward-rightward and forward-leftward,* $p<0.002$.

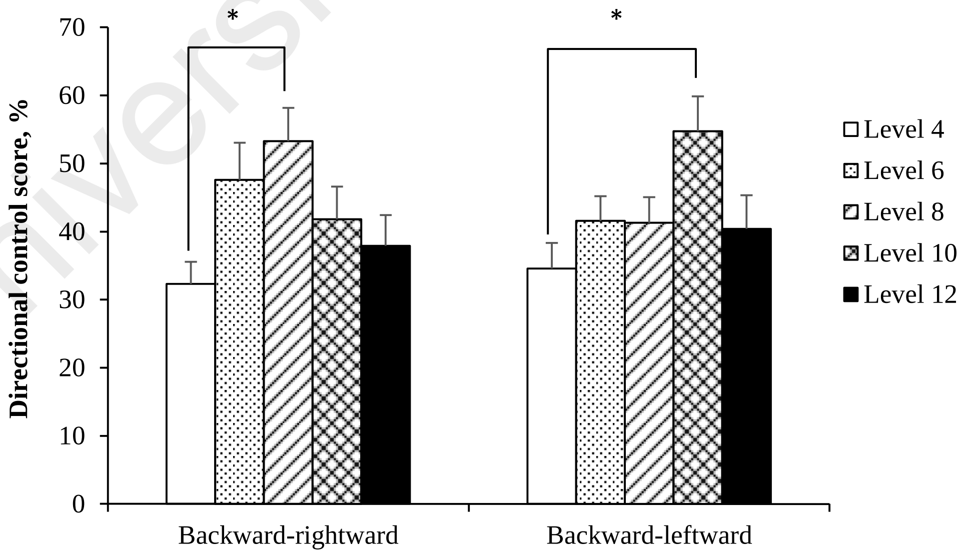

Figure 6.4: The directional control score of eight predetermined target direction for different dynamic stability levels in backward-rightward and backward-leftward, * $p<0.002$

6.3.2 Muscle Activity

Table 6.2 presents the changes of muscle activity in different stability levels for TTAs. With the available number of participants in this study, the statistical analysis indicated a significant main effect of the changes of EMG activity levels ($F_{1.84,16.54} = 4.94$, $p = 0.023$). The multiple comparisons revealed a significant difference of EMG activity between left-right biceps femoris ($p = 0.025$), right rectus femoris–right biceps femoris ($p < 0.001$), and left rectus femoris–right biceps femoris ($p = 0.005$). No significant differences of EMG activity were found between stability levels. No interaction effect between stability level and muscle group was observed.

Table 6.2: Statistical findings of all the recorded muscles (mean of normalized RMS ± SD) in different stability levels among TTA.

Muscles	Stability level									
	Level 4		Level 6		Level 8		Level 10		Level 12	
	Mean ± SD	95% CI	Mean ± SD	95% CI	Mean ± SD	95% CI	Mean ± SD	95% CI	Mean ± SD	95% CI
Rectus Femoris										
Left	1.815 ± 1.075	1.243-2.388	1.648 ± 0.987	1.122-2.173	1.679 ± 0.915	1.192-2.166	2.017 ± 1.411	1.235-2.798	2.077 ± 1.499	1.247-2.907
Right	1.694 ± 0.845	1.244-2.145	1.674 ± 0.898	1.195-2.153	1.849 ± 1.089	1.269-2.429	1.907 ± 0.938	1.387-2.426	2.236 ± 1.447	1.434-3.038
Cohen's d	0.13		0.03		0.17		0.09		0.11	
p-value	0.726		0.937		0.635		0.803		0.769	
Biceps Femoris										
Left	2.216 ± 2.288	0.996-3.435	2.271 ± 2.418	0.982-3.559	2.242 ± 2.541	0.888-3.596	2.553 ± 2.627	1.098-4.008	3.442 ± 4.236	1.097-5.789
Right	1.289 ± 0.948	0.784-1.794	1.109 ± 0.389	0.902-1.317	1.088 ± 0.211	0.976-1.201	1.084 ± 0.154	0.999-1.169	1.072 ± 0.095	1.247-2.907
Cohen's d	0.53		0.67		0.64		0.79		0.79	
p-value	0.145		0.067		0.080		0.039*		0.039*	
Medial Gastromedial										
Left †	1.681 ± 1.906	0.216-3.146	1.588 ± 1.720	0.267-2.910	1.409 ± 0.994	0.645-2.173	1.732 ± 1.158	0.842-2.623	1.543 ± 1.278	0.560-2.525
Right †	2.018 ± 1.528	0.604-3.431	2.098 ± 1.722	0.506-3.691	1.828 ± 1.236	0.685-2.972	1.694 ± 1.117	0.522-2.866	1.566 ± 0.684	0.848-2.284
Cohen's d	0.19		0.30		0.37		0.03		0.02	
p-value	0.709		0.566		0.464		0.950		0.968	

* Indicate significant changes between left and right side of the muscle group ($p<0.05$).
† There are left amputated ($n = 9$) and right amputated ($n = 7$) for medial gastrocnemius muscle groups.

6.4 Discussion

This study aimed to investigate the changes of directional stability and muscle activity in 5 different dynamic stability levels for TTAs. The findings indicate that reduction of the CoP displacement was accompanied with the increased of support surface perturbation. The study also found that EMG activity of lower limb was significantly different between the left and right side for the biceps femoris muscle group. There was no specific association between directional control and lower limb EMG activity.

In the present study, the findings showed that it is more challenging for the TTAs to control body leaning in ML direction than the AP direction. The findings also revealed that the forward direction is the easiest body leaning direction that allows a subject to achieve the goal in this study. Generally, young adult tends to exert a slightly greater postural sway in the AP direction in a quiet stance (Ku et al., 2012a). The ankle joint serves to control the postural sway in AP and ML direction for small perturbation. With the loss of anatomical ankle joint, the mechanism to control postural balance will be differ from the norms. In the study of Curtze et al. (2012), they reported that the sound ankle of TTA will fully compensate for the active ankle moment in the prosthetic limb for an induced forward fall situation. The role of the ankle joint in sideward direction will decline as postural stabilization is maintained more by hip strategy. Approximately 61% of the load distribution relies more on the sound hip joint for an induced ML fall (Curtze et al., 2012). However, a long period of weight loading on the intact leg would increase the prevalence of lower back pain and osteoporosis in the residual leg in amputees (Ventura, Klute, & Neptune, 2011). Similarly, a threatening situation will also cause muscle co-contraction and increased support surface perturbation, further decreasing the CoP amplitude and increasing the mean frequency in the sideward direction (Carpenter, Frank, Silcher, & Peysar, 2001). The postural sway for directional stability was gradually increased when the platform tilt angle inclined from level 8 (7.2º) to level 4 (14.4º) for the moving platform, while the postural sway at level 10 (3.6º) is slightly smaller than level 8. This suggests that the flexible slope degree of the moving platform at level 10 might limit the control of body leaning and the tilt angle of 3.6º might not be suitable to evaluate the perturbation in balance during a dynamic LoS test.

Furthermore, the results of the present study showed that the TTAs tend to rely more on their intact leg for the static platform condition (Level 12). In line with previous studies, the mean displacement and root mean square of CoP were greater for the side of the intact leg for TTAs during stance (Quai, Brauer, & Nitz, 2005; Vrieling et al., 2008). Quai, Brauer, and Nitz (2005) revealed the CoP excursion of the intact leg for AP and ML direction was greater than the prosthetic limb during both eyes open and eyes closed condition. The removal of part of the lower limb due to amputation will also reduce the proprioceptive foot sensation and lower limb muscle strength in the residual leg, further hindering the confidence level. In order to compensate for the proprioception deficit due to the lack of sensory input at the residual leg, the body would control postural balance more to rely on the hip musculature for the residual leg (Isakov, Mizrahi, Ring, Susak, & Hakim, 1992).

Despite of asymmetry of postural sway between the intact leg and residual leg for all stability levels in the current study, the findings showed the muscle activity difference of side-to-side lower limb for the biceps femoris muscle groups. The rectus femoris muscle were not active for dynamic stance control. The difference in biceps femoris activity indicates that the amputee tends to exert higher biceps femoris muscle activation at their intact leg. This is consistent with previous studies that the intact leg for lower limb amputees has greater stance time, ground reaction force and joint moment (Isakov, Keren, & Benjuya, 2000; Ventura, Klute, & Neptune, 2011). Although the muscle activity does not seem to be accompanied with platform stiffness, there is a trend that the biceps femoris and medial gastrocnemius muscle activity are gradually increasing and approximately equal with the left biceps femoris when the platform stiffness declines. Additionally, the study of de Freitas et al. (2009) also reported that the hip abductor was responsible for the sideward perturbation. For the hip abductor, the gluteus maximus is involved in the forward and backward pelvis tilt, while gluteus medius is involved in the sideward pelvis tilt for dynamic stabilization. Jandric (2007) found that the rehabilitation program with prosthetic limb will improve the isometric hip muscle strength for TTAs.

There are several limitations in the current study. Firstly, the sample size of the study group was considered small and limited to TTAs with a sedentary lifestyle. Hence, the findings may not be generalized to experienced lower limb amputees with higher activity levels. Secondly, the foot position was not standardized for all subjects. In order to mimic dynamic standing in real-life, the subjects are instructed to stand according to their comfort standing posture. However, dynamic standing applies to real-life simulation activity such as standing in a moving bus, and this study only could evaluate the ability of amputees handle certain types of balance perturbation. Thirdly, as most of the previous studies found that no functional asymmetry for dominant and non-dominant lower limb in dynamic postural stability for healthy adults (Hoffman, Schrader, Applegate, & Koceja, 1998; Wikstrom, Tillman, Kline, & Borsa, 2006), the effect of limb dominance on the intact limb and prosthetic limb was assumed to be negligible in this study. The current study is of particular interest to researchers or physiotherapists who are interested in postural control of dynamic standing among TTAs. Future studies may consider the effect of limb dominance on the balance control for lower limb amputees in the dynamic standing.

The current chapter found that the reduction of platform stiffness would increase the CoP displacement of the directional control in a dynamic stance. In addition, the findings found that TTAs generated a higher muscle activation at their intact leg than their amputated leg only in the biceps femoris. The rectus femoris did not have significant changes towards the inclination platform stiffness. Therefore, it is concluded that the dynamic stance from 0º to 14º may only affect directional stability and the muscle activation of the lower limb may not have significant changes. Further studies are suggested to examine the functional relevance of limb dominance on dynamic standing among lower limb amputees. The dynamic stance will affect the ability of postural balance, while the lower limb muscle activation will remain unchanged. The following chapter will examine on the possible anthropometric variables which might affect postural control on transtibial amputees.

CHAPTER 7:
THE CHANGES OF BALANCE CONTROL FOR TRANSTIBIAL AMPUTEE IN STATIC AND DYNAMIC STANDING

7.1 Introduction

The balance ability is a fundamental skill that used to perform goal-directed movement and desired activities of daily living. An estimated of 185,000 people undergo limb amputation each year in the United States, and a majority of amputations occur in lower limbs (Owings & Kozak, 1998). One of the major balance-related concern is the risk of falling and approximately 52.4% of community-dwelling elderly amputees will experience fall during their daily activities (Miller, Speechley, & Deathe, 2001).

The removal of a body segment and functional impairment in lower extremities of amputees interfere with system integration, and generates changes in the ground reaction force and CoP shifting pattern, thereby affecting postural control. Proficiency in an amputee's postural stability is an indication that they have been well-adapted to the environment through adaptation of a new balance recovery strategy. Several studies reported that lower extremity amputees have a higher CoP excursion, CoP velocity and ground reaction force than normal individuals (Buckley, O'Driscoll, & Bennett, 2002; Duclos et al., 2009; Duclos et al., 2007; Vrieling et al., 2008). The increased of CoP excursion and velocity would result in balance instability and increase the falling risk.

To date, many studies have investigated the postural control of amputees in the aspects of gait pattern, intervention and rehabilitation. However, limited studies have focused on the regulation of static balance among LEAs. Static balance control serves as a balance indicator of dynamic control via the assessment of postural sway. The body segment displacement, muscle activity, and movement pattern of the CoM and CoP are the three main aspects used to evaluate human standing (Balasubramaniam & Wing, 2002).

Several studies showed that a majority of amputees tend to shift their body weight to the non-amputated leg than the amputated side (Duclos et al., 2009; Hlavackova et al., 2011; Quai, Brauer, & Nitz, 2005). Curtze et al. (2012) found that the non-amputated leg has a higher joint moment during directional fall-induced testing. Additionally, the LEAs have poor balance in both static and dynamic conditions compared to normal individuals (Buckley, O'Driscoll, & Bennett, 2002). LEAs with poor balance are at increased risk of falling and muscle weakness. Poor balance is also related to a long period of exposure to the prosthetic limb. Due to the higher repetitive loading forces in the intact limb, the degeneration of the intact joint and destruction of cartilage, decreased of muscle mass lead to secondary complication such as osteoarthritis, osteoporosis, back pain and musculoskeletal problems (Dequeker, Aerssens, & Luyten, 2003).

BMI is a type of calculation of relative weight and obesity, where the body weight is divided by the square meter of body height (kg/m^2) (Keys et al., 1972). It attempts to estimate the amount of fat tissue based on the body height and weight. Obesity would deteriorate the balance ability and diminish the anterior-posterior stance stability in healthy adolescents. However, the calculation of BMI does not take into account muscle mass, bone density, or missing limbs. Therefore, it remains unclear whether such changes of BMI would influence on the postural stance stability for LEAs. Measurement of anthropometric variables could be a method to identify the changes in body composition towards the amputees' postural stance balance. A more comprehensive understanding of the association between postural balance and BMI would help in developing a better intervention program to improve the postural balance of LEAs following lower extremity amputation. Therefore, the aim of this study was to investigate the possible balance-related factor in static and dynamic stance and to determine the effect of stance posture on postural stability.

7.2 Methods

7.2.1 Subjects

Eighteen unilateral TTAs and 18 able-bodied controls were recruited in this study. The reasons for amputation were trauma (*n*=10) and vascular disease (*n*=8). The inclusion criteria for LEAs were aged over 35 years, experienced lower extremity amputation at least 1 year earlier, the use of a prosthetic on a daily basis for at least 6 months, and able to stand without walking aids for at least 30 minutes. Control subjects were excluded if they had suffered from any prior lower extremity injury, neurological, vestibular impairment, or balance disorders. All procedure were approved by the Medical Ethics Committee (MEC 895.7) and all subjects signed written informed consent forms to participate in the study. The participants were divided into two BMI categories: normal weight (≤ 24.99 kg/m^2) and overweight (≥ 25.00 kg/m^2). All assessments were measured by a trained research assistant in the Body Performances Laboratory. All participants underwent a comprehensive assessment protocol. The participants were provided instructions for all assessments.

7.2.2 Clinical Functional Assessment

Before the postural stability test was commenced, a series of clinical assessment were conducted for TTAs and healthy control groups. Both groups were assessed with Fall Efficacy Scale (FES), skinfold measurement, two-point discrimination test, vibration sensation and Q-angle measurement. The functional capability for TTAs and healthy controls were assessed with the Amputee Mobility Predictor with the Use of Prosthesis (AMPPRO) and Berg Balance Scale respectively.

7.2.2.1 Berg Balance Scale

The balance ability for able-bodied controls was evaluated using the Berg Balance Scale (Berg & Norman, 1996). The Berg Balance Scale is a commonly used clinical functional assessment by doctors and physiotherapists. The total of the Berg Balance score is 56. The items of the Berg Balance Scale are shown in Appendix A.4.

7.2.2.2 Fall Efficacy Scale

The level of falling concern during daily functional activity was assessed using the FES. The FES has 16 items used to assessed the functional activity on 4 level scale based on the level of task consideration ranging from 1 (not consent) to 4 (very consent) (Appendix A.5). The range of total scores is from 16 to 64 with the higher score indicating better skill.

7.2.2.3 Amputee Mobility Predictor with the Use of Prosthesis

The functional capability of amputees in ambulation was determined using AMPPRO. It consists of 21 items that generates scores based on the different movement abilities (Appendix A.6). The sum of the score yields an AMPPRO score ranging from 0 – 64. A higher AMPPRO score represents a good mobility level. Gailey et al. (2002) reported that AMPPRO is a reliable assessment with inter-rater reliability (r = 0.99) and intra-rater reliability (r = 0.96).

7.2.3 Skinfold Measurement

The estimation of body composition was assessed by a Harpenden skinfold caliper (Baty International, West Sussex, UK). Based on the assumption that subcutaneous fat appears as total fat, the skinfold caliper measured the skinfold thickness at 7 sites (triceps, chest, subscapular, mid axillary, suprailiac, abdominal, and mid-thigh) to the nearest 0.1 mm (Weiner & Lourie, 1969). All measurements were taken on the right side of the body. Three non-consecutive replicates were measured and were subsequently averaged. The sum of the 7-fold measurement was calculated and the equations from Jackson and Pollock (1978) were used to calculate the body fat % (Jackson & Pollock, 1978; Jackson, Pollock, & Ward, 1980).

7.2.4 Two-point Discrimination Sensitivity Test

An aesthesiometer (Lafayette Instrument Co., IN, USA) was used to measure the tactile sensitivity of the foot sole, trans-metatarsal, and mid-heel region. Participants were asked to describe the number of touch points, while the aesthesiometer was gradually decreased from a maximum distance and stopped when they could not differentiate the two touch points on their foot. Three trials were performed and the average was recorded in mm. All measurements were taken on the right side of the body.

7.2.5 Vibration Sensation

A 128Hz tuning fork was used to assess the vibration sensitivity at the first -head metatarsal and medial-malleolus of the foot. The time began to count when the tuning fork was placed on the subjects' foot and stopped when they felt that the vibration of tuning had stopped. All measurements were taken on the right side of the body. The period of vibration for 3 trials was recorded in seconds.

7.2.6 Q-angle Measurement

The Quadriceps or Q-angle is an index of vector for a combination of patellar tendon and extensor mechanism that indicates the dysfunction of the patellofemoral joint (Smith, Hunt, & Donell, 2008). It was measured by a lengthened-arm 360° universal goniometer. Participants were positioned standing with both feet parallel and all measurements performed by the same investigator. The location of tibial tuberosity, anterior-superior illiac spine (ASIS) and middle of the patella were marked. The pivot of the goniometer was placed on the center of patellar. The goniometer's arms were placed along the line connecting tibial tuberosity-middle of patellar and ASIS-middle of patellar. The angle that intersected the 2 anatomical axes of the frontal plane was identified as the Q-angle. The measurement for TTAs and controls were taken from the intact side and right side respectively.

7.2.7 Postural Balance

The balance performance was measured by the BBS (Biodex Medical System, Inc. NY, USA). It is used to measure the displacement of CoP, with a circular platform which allows up to 20° of platform tilt in a 360° range of motion. The data were measured by 4 strain gauges sampled at a rate of 20 Hz. OSI, MLSI and APSI were the measured balance variables for the postural stability test. All indices were calculated as the standard deviation of CoP sway around the origin zero point and the units were recorded in degrees. MLSI was measured from the horizontal displacement along medial-lateral axes as x-direction, while APSI was assessed from the vertical displacement along the anterior-posterior axes as y-direction. The combination of horizontal and vertical displacement generated the OSI. Cachupe et al. (2001) found that the reliability for 3 trials was 0.83 (OSI), 0.81(MLSI) and 0.73 (APSI).

Postural balance was assessed using a postural stability test for static stance (static level) and dynamic stance (level 6) for a period of 30 s. For the control group, subjects were stood on the circular platform with shoes and their arms placed along their side. For the amputee group, they were required to stand at the platform in their prosthetic limb with shoes. All subjects were instructed to stand comfortably and focus straight ahead on the display screen. They were instructed to adjust their foot placement into a comfortable posture that allows them to maintain the pointer at the origin point. The foot placement was required to remain constant once the test commenced. The position of the foot placement was recorded. Two practice trials were performed in order to minimize the learning effect. Subjects were required to maintain the pointer at the origin point throughout the 30 s test. Three trials were performed with 10 s of resting between each trial. The order of testing was randomized.

7.2.8 Statistical Analysis

All the statistical analyses were performed using SPSS 19.0 for Window (Version 19, IBM Corp., Armonk, NY). The normality that was assessed by Shapiro-Wilk test showed that all the data were normally distributed ($p > 0.05$). A mixed factor repeated measures ANOVA, 2 x 2 x 3 (BMI group [normal – weight vs. overweight] x stance posture [static vs. dynamic] x relative stability indices [OSI vs. APSI. Vs. MLSI]) was employed to analyze the effect of stance condition on postural stability. To assess the tested group differences, a 2 x 3 (tested group x stability indices) ANOVA was determined for each stance condition. The Bonferroni *post-hoc* analysis was computed for multiple comparisons. A Mann-Whitney test was conducted to compare the difference of foot vibration sensation, two-point discrimination sensitivity and Q-angle measurement between the amputee and control groups. Binary logistic test was used to analyze the sensitivity of all variables among TTA and controls. The spearman rank correlation was used to determine the relationship between all anthropometric variables and functional assessment for amputee and control groups. Stepwise multiple regression analysis was computed to predict the postural control of TTAs from age, height, BMI, vibration sensation, two-point discrimination at foot sole, and Q-angle at static stance and dynamic stance. Relative stability was used in the analysis, by normalizing the data with respect to body weight in order to avoid potential misinterpretation of data and reduce the absolute stability differences (Winter, 1995). All the data were expressed as mean and standard deviation (mean ± S.D). The level of significance was set at $p \leq 0.05$.

7.3 Results

The demographic characteristics for the amputee and control groups are listed in Table 7.1. The FES score is similarly for TTA and control groups. There are no significant main interactions between the tested group, stance conditions and BMI groups. The descriptive statistics of the mean relative stability score for the static and dynamic stance are presented in Table 7.2 and Table 7.3. The BMI, sum of skinfold, and body fat % were found to be significant between the normal weight and overweight group.

In the TTA group, the ANOVA exhibited a significant interaction between BMI group, stance posture and stability indices ($F_{2,32} = 12.31$, $p < 0.001$). Statistical analysis revealed significant main effects of stance posture ($F_{1,16} = 6.443$, $p = 0.022$) and stability indices ($F_{1.45,23.22} = 77.93$, $p < 0.001$) on the postural stability. It indicates that the mean relative stability scores differed between static and dynamic stance (Figure 7.1). The interaction effect of stance posture x BMI ($p = 0.001$) and stance posture x stability indices ($p = 0.013$) were observed in the analysis. In the normal weight group, a higher mean static relative stability score was found, while the overweight group showed a higher mean dynamic relative stability compared to the static stance.

Table 7.1: The physical characteristic for transtibial amputee and control groups.

Variables	Amputee (n=18)	Control (n=18)	p-value
Age (years)	50.1 ± 6.4	49.3 ± 6.1	0.712
Body weight (kg)	73.7 ± 16.7	68.9 ± 14.9	0.368
Body height (cm)	166.7 ± 7.0	160.2 ± 10.4	0.034*
Body mass index (kg/m²)	26.4 ± 5.1	26.9 ± 5.5	0.792
Body circumference			
Waist (cm)	95.1 ± 13.4	88.0 ± 10.6	0.090
Hip (cm)	101.6 ± 9.7	102.0 ± 9.7	0.908
Waist-Hip ratio	0.93 ± 0.09	0.87 ± 0.06	0.006
Year since amputation (years)	6.9 ± 5.5	-	-
AMPPRO score	29.4 ± 9.4	-	-
Berg Balance score	-	55.9 ± 0.3	-
Fall Efficacy score	31.6 ± 8.0	27.9 ± 8.4	0.189

* $p<0.05$ for the comparison between amputee and control groups.

Table 7.2: Comparison of stability between BMI groups for transtibial amputees.

Variables	Amputee	
	Normal weight (n=8)	Overweight (n=10)
Reason of Amp. (Trau/Diab)		
Body Mass Index (kg/m²)	22.35 ± 2.20†	29.62 ± 4.29
Σ Skinfold (mm)	166.68 ± 52.69†	244.66 ± 76.74
Body Fat (%)	25.39 ± 8.52†	32.66 ± 5.81
*Static standing (°)**		
OSI	14.57 ± 10.24†	5.92 ± 2.70
MLSI	5.63 ± 6.46‡	2.77 ± 1.29
APSI	8.73 ± 8.66	4.43 ± 2.19
*Dynamic standing (°)**		
OSI	12.65 ± 5.04	19.88 ± 7.78
MLSI	6.32 ± 3.19	10.93 ± 6.37
APSI	7.80 ± 3.47	11.37 ±5.41

* Values with expression of x^{-3}.
† $p<0.05$: significant differences when compared with normal weight and overweight groups.
Amp: amputation; Trau: traumatic; Diab: diabetes. All values are expressed as mean ± standard deviation.

Table 7.3: Comparison of stability between BMI groups for control group.

Variables	Control	
	Normal weight (*n*=8)	Overweight (*n*=10)
Body Mass Index (kg/m²)	22.09 ± 2.45†	30.68 ± 4.09
Σ Skinfold (mm)	164.85 ± 37.69†	252.61 ± 59.20
Body Fat (%)	27.25 ± 5.87†	37.20 ± 6.00
*Static standing (º)**		
OSI	7.27 ± 2.51†	6.19 ± 3.17
MLSI	3.35 ± 1.84	2.98 ± 1.44
APSI	5.19 ± 1.89	4.46 ± 2.61
*Dynamic standing (º)**		
OSI	12.70 ± 4.96	17.18 ± 6.05
MLSI	7.22 ± 3.17	8.04 ± 3.03
APSI	8.22 ± 2.72	10.49 ±3.56

* Value with expression x^{-3}.
† p<0.05: significant differences when compared with normal weight and overweight groups.
All values are expressed as mean ± standard deviation.

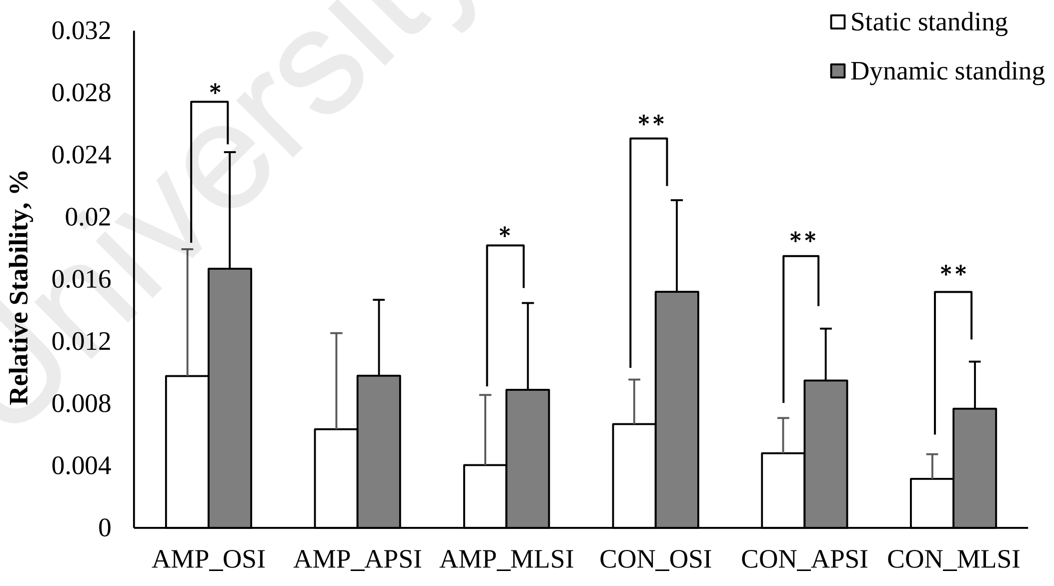

Figure 7.1: The difference of relative stability score for amputee and control groups.* p<0.05; **p<0.001

For the healthy control, a significant interaction between BMI group, stance posture and stability indices ($F_{2,32} = 6.527$, $p = 0.004$) was observed. The ANOVA demonstrated significant main effects of stance posture ($F_{1,16} = 57.55$, $p < 0.001$) and stability indices ($F_{1.42,22.79} = 72.57$, $p < 0.001$). The comparison between stance postures showed significant differences for all stability indices. The OSI, APSI, and MLSI for dynamic stance was significantly greater than static stance. Statistical analysis revealed an interaction effect between stance posture x BMI ($p = 0.047$) and stance posture x stability indices ($p < 0.001$). Both the normal weight and overweight groups exhibited a greater dynamic relative stability score. A significant difference between OSI–MLSI were observed in amputee ($p = 0.031$) and control ($p < 0.001$).

The ANOVA showed no differences in relative stability between amputees and control groups in both stance postures. The results of the ANOVA showed a significant main effect for both tested groups ($F_{1,102} = 3.93$, $p = 0.05$) on postural stability ($F_{2,102} = 8.24$, $p < 0.001$). Generally, the amputee and control groups had a significantly higher mean APSI score than MLSI score (Figure 7.1).

Statistical analysis indicated that only the distance of two-point discrimination sensitivity at mid-foot was significantly wider in TTAs compared with controls ($p = 0.001$, Table 7.4). The duration of vibration measurement for TTAs were shorter compared with controls at the first metatarsal head and medial malleoli. No significant difference of Q-angle was found between the intact leg of TTAs and right leg of TTAs. A significant correlation was found for body weight with the two-point discrimination sensitivity at trans-metatarsal ($r = –0.787$, $p < 0.001$), mid-foot ($r = –0.784$, $p < 0.001$), and heel ($r = –0.570$, $p = 0.014$) (Figure 7.2). The BMI also showed a significant correlation with trans-metatarsal ($r = –0.752$, $p < 0.001$), mid-foot ($r = –0.826$, $p < 0.001$), and heel ($r = –0.653$, $p = 0.003$) (Figure 7.3). The multiple regression analysis does not display a significant correlation of postural stability with the two-point discrimination sensitivity, vibration sensation or Q-angle at static stance and dynamic stance. Table 7.5 presented the sensitivity for all tested variables in static and dynamic stance conditions.

Table 7.4: Comparison of two-point discrimination sensitivity, vibration sensation and Q-angle measurement of TTA and healthy control.

Variables	TTA	Control	Cohen's *d*	*p*-value
Two-point discrimination (mm)				
Trans-metatarsal	1.36 ± 0.61	1.17 ± 0.34	0.39	0.540
Mid-foot	1.50 ± 0.53	0.99 ± 0.24	1.24	0.001
Heel	1.68 ± 0.82	1.39 ± 0.61	0.40	0.362
Vibration (s)				
First metatarsal head	12.62 ± 4.57	17.59 ± 4.59	1.09	0.006
Medial malleol	12.07 ± 4.38	17.09 ± 4.66	1.11	0.005
Q-angle (º)				
Intact/Right leg	15.5 ± 4.0	15.8 ± 3.6	0.08	0.680

All value presented as mean ± SD.

Table 7.5: The sensitivity analysis for anthropometric variables for TTA and controls.

Variables	Sensitivity (%)
Age (years)	47.06
Body weight (kg)	52.17
Body height (cm)	64.00
Body mass index (kg/m^2)	55.17
Waist to hip ratio	68.42
Body Fat (%)	61.90
Postural control	
Static stance	64.29
Dynamic stance	53.33
Two-point discrimination (mm)	
Trans-metatarsal	54.17
Mid-foot	23.08
Heel	39.13

Table 7.5, continued: The sensitivity analysis for anthropometric variables for TTA and controls.

Variables	Sensitivity (%)
Vibration (s)	
First metatarsal head	23.08
Medial malleol	46.15
Q-angle (º)	
Intact/Right leg	29.63

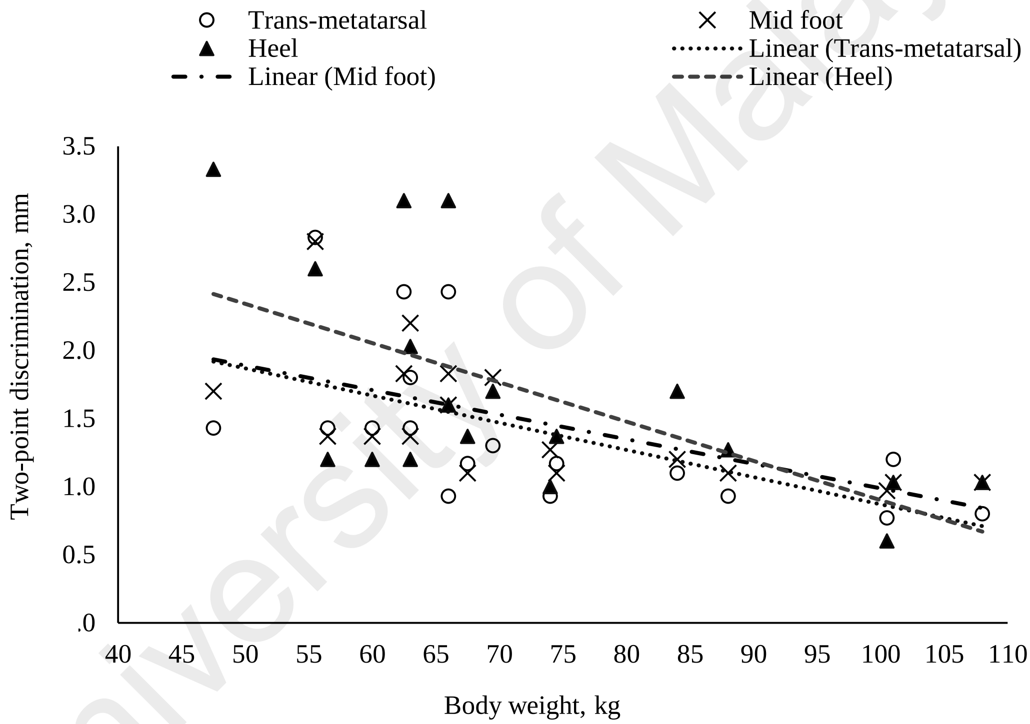

Figure 7.2: The association of body weight with the three measurement sites of two-point discrimination sensitivity test (Trans-metatarsal, mid-foot and heel).

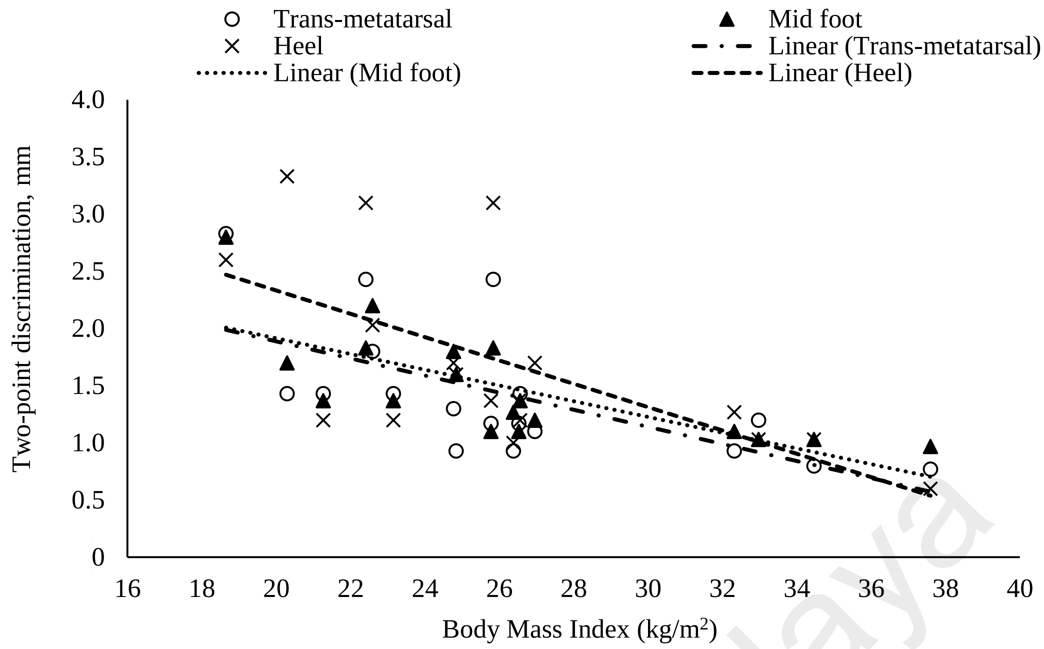

Figure 7.3: The association of BMI with the three measurement sites of two-point discrimination sensitivity test (Trans-metatarsal, mid-foot and heel).

The Spearman rank correlation revealed a weak negative correlation for BMI–static relative stability (r = –0.471, p = 0.049) for the amputee group. The controls also showed a moderate correlation was observed for BMI–static relative stability (r = 0.506, p = 0.0032) and body fat %-static relative stability (r = 0.536, p = 0.022). No correlation between BMI and dynamic relative stability were found. Besides, there is a significant correlation of body fat % and BMI in amputee (r = 0.851, p < 0.001) and control groups (r = 0.911, p < 0.001) (Figure 7.4).

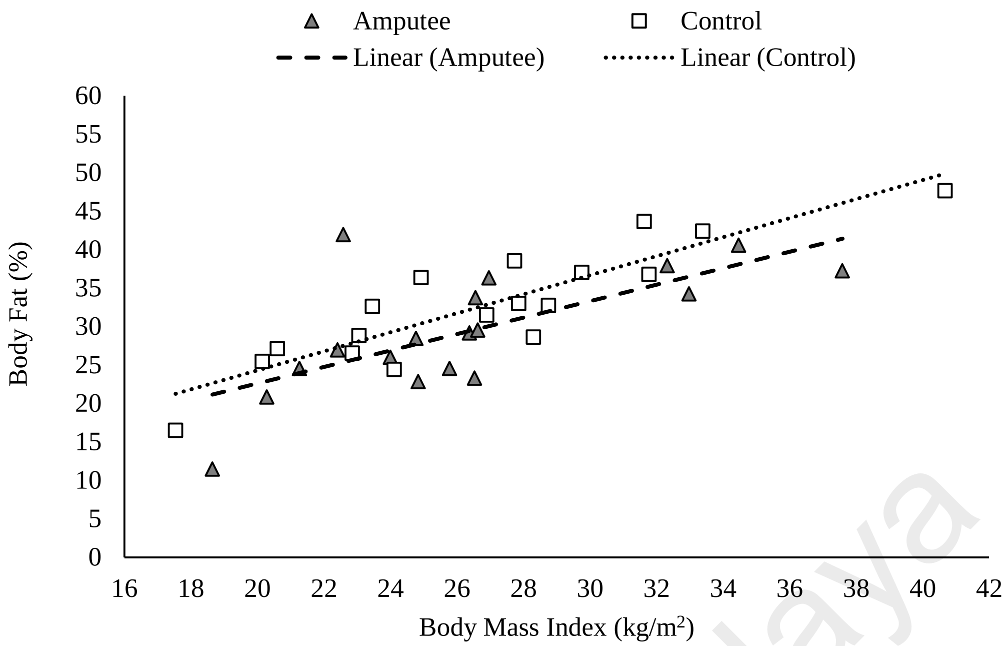

Figure 7.4: The association between BMI and body fat % in static stance condition.

7.4 Discussion

The primary purpose of this study was to investigate the relationship of BMI categories and stance stability among unilateral transtibial amputees and the changes of stance posture towards balance control. The findings showed that the increase of BMI was negatively correlated with static relative stability among transtibial amputees and no correlation was found in dynamic stance. Interestingly, the normal weight group tend to exhibit slightly greater static relative stability in static stance, while the overweight displayed a greater dynamic relative stability compared to normal weight in both the amputee and control groups. There are no significant differences of relative stability between amputee and control groups. Additionally, no significant differences of relative stability between OSI–APSI and APSI–MLSI for transtibial amputee and control group were found.

The findings demonstrated that the overweight amputee tends to display a greater postural sway than a normal weight amputee in the dynamic stance, as the differences of dynamic relative stability for normal weight and overweight are nearly significant. Recent studies have reported that increased BMI for the able-bodied exhibited a faster oscillation, and further deteriorated mobility performance with a greater postural sway in order to maintain stance balance (Dutil et al., 2013; Hergenroeder, Wert, Hile, Studenski, & Brach, 2011; Ku et al., 2012a). In addition, the changes of stance posture from static to dynamic have induced a reduction of CoP sway in normal weight amputees and increased CoP sway in overweight amputees. These results were in line with the changes of relative stability of the control group. It is suggested that the distribution of adipose tissue caused the increased of CoP sway for the overweight in dynamic stance. The excessive adipose tissue would modify the body geometry in either the android shape or gynoid shape. The increase of abdominal fat would contribute to the anterior shift of CoM (Li & Aruin, 2009). The CoM alteration may impact on the CoM velocity, magnitude of ankle torque and range of motion for body trunk during the dynamic stance.

According to the current findings, there is no difference of CoP sway in the AP and ML axes for the amputees or control groups in either stance posture. Previous studies have elucidated that LEAs generate a greater AP sway area than the able-bodied during a single-leg-raising stance posture, while no side effect was found in the ML axis (Mouchnino et al., 1998; Viton et al., 2000). Controversially, Buckley, O'Driscoll, and Bennett (2002) found the LEAs displayed a greater AP and ML sway compared to the able-bodied in the dynamic stance control. This may relate to the contribution of ankle joints in balance control. The loss of ankle joint had restricted the AP sway and results in the asymmetry of weight distribution among amputees (Duclos et al., 2009; Quai, Brauer, & Nitz, 2005). As the ankle joints regulate more on the anterior-posterior axis, the extended knee joints caused the activation of the hip to compensate for the limited regulation in lateral movement. The majority of the amputees are more likely to shift their weight-bearing to the intact side. Quai, Brauer, and Nitz (2005) reported that the intact limb displayed a greater CoP sway in the AP and ML axes. The tendency of the asymmetrical weight distribution would apply additional forces to the joints, ligament, tendons, and muscle. Kalbaugh et al. (2006) revealed that functional outcomes such as prosthetic usage, ambulation level, survival, and independent living status did not impact on the postural control of obese dysvascular amputees. Furthermore, the asymmetrical weight distribution on the intact side may alter the natural bone alignment and induces muscle weakness.

Compared with the changes between relative stability between stance postures, the dynamic relative stability of overweight TTAs was increased significantly. This may reflect on the impact of the removal of lower-extremity body segment towards balance control (Buckley, O'Driscoll, & Bennett, 2002; Mayer et al., 2011). In particular, LEAs performed greater postural sway compared to the able-bodied in dynamic stance. It is possible that the anatomical loss of ankle joint would alter the postural stability by regulating the musculoskeletal leg alignment. The hip, knee, and ankle joints are naturally utilized to preserve postural stability. The removal of ankle or knee joints results in the adjustment of habitual control, and development of a new balance adaptation strategy (Vrieling et al., 2008). The loss of sensory modalities also would contribute to the functional loss, and increase the risk of ulceration and second leg amputation (Llewelyn, 2003). This highlights that the impact of proprioception loss on the vibration, light touch, joint position, pain, and temperature senses could increase risk of falling. A repeated overloaded intact limb over a long period of time may also result in inflammation, knee and low back pain, muscle injury or symptomatic osteoarthritis (Norvell et al., 2005).

Underlying the significant negative correlation with increased BMI in static stance, there are weak, but not significant correlations observed in dynamic stance. The weak correlation of relative stability reflects the increased postural sway with increased BMI. Moreover, the current findings showed that there is a strong association between BMI and body fat % (skinfold measurement). The high sensitivity and strong association may indicate the reliability of BMI for transtibial amputees in dynamic stance. Zanovec, Lakkakula, Johnson, and Turri (2009) reported high physical activity is associated with a lower body fat %, fat mass and higher lean tissue.

The findings revealed that there is no relationship of postural control with foot sensation (two-point discrimination and vibration tests). However, the distance of two-point discrimination sensitivity at mid-foot was increased compared with the control group. Additionally, the shorter vibration sensation for TTAs reflects the loss of sensation of the foot sole. It is speculated that a continuous period of load distribution at the intact limb would increase the sensory loss of the sole of foot. As the Q-angle indicated the patellofemoral joint deformation, the present result showed that TTAs will not suffer patellofemoral dysfunction after a prolonged time of wearing a prosthetic limb. Furthermore, the results showed that foot sensation was reduced with the increase in body weight and BMI. This had indirectly highlighted that the diabetic TTAs are commonly associated with obesity (Klein et al., 2004).

There are several limitations which should be considered in the current study. Firstly, this only evaluated stance balance for a small sample size of sedentary LEAs aged 25-60. Secondly, the participants were LEAs with at least 1 year experienced in wearing a prosthetic. The current findings cannot apply for the new prosthetic user, as Mayer et al. (2011) found that first-fitted users have a higher load distribution and CoP excursion than experienced users. In addition, different types of prosthetic might slightly affect the result. However, as they are skilled users, it is assumed that they are well adapted to their prosthetic in daily activities. The study of amputees' postural control in dynamic standing has provided valuable information that could help in the development of rehabilitation programs for amputee.

In summary, the findings of current chapter highlight that relative stability for LEAs is negatively correlated with increased of BMI in static stance. Although there is no significant correlation between dynamic relative stability and BMI in either tested group, there is a tendency of overweight amputees to exhibit greater postural sway than normal weight amputees in dynamic stance. In addition, no differences of static and dynamic relative stability were found between the amputee and able-bodied groups. An association of body weight and BMI with the two-point discrimination sensitivity at mid-foot and heel was found. None of the variables such as age, weight, height, Q-angle, body mass index, skinfold measurement and foot sensations were found to be a predictor for postural control among TTAs. The current findings have eliminated the related variables as possible balance-related factors among transtibial amputee for future research.

CHAPTER 8:
CONCLUSION AND RECOMMENDATION

Postural stance plays a significant role as the locomotion basis of performing daily tasks. Postural control is regulated through the integration of various systems in the body. The failure to maintain body balance has been related with increased falling risks, which would result in injury, bone fracture or even death. However, there are gaps in how the balance-related factors could impact on postural stance control among adults and lower limb amputees. Therefore, all chapters were summarized based on the research aims and study limitations respectively.

In order to assess the postural activity among healthy young adults according to their BMI classification and gender during bipedic stance and unipedic stance, the findings in Chapter 3 revealed that the static relative stability gradually increased as the body mass index declined. In addition, there is a tendency in females to exhibit greater center of pressure displacement compared to males in certain body mass index classifications. The impact of leg dominance was neglected in this study. The learning ability of an individual would also impact on the study findings. Moreover, all participants were from the same geographical location.

In Chapter 4, the changes of postural stability in bipedic stance and stance with toe-extension among young adult were determined. The postural control was not influenced by minor changes of the foot support area. The postural performance in bipedic stance and stance with voluntary toe-extension were similar. There are no gender effects on static postural stability among young adults. These current findings were only applied for active toe-extension standing. The frequent high-heel wearer was excluded from the current study. The participants were from the same geographical location with sedentary lifestyles.

Moreover, the muscular response on the postural activity during the support surface perturbation among middle-aged adult were investigated in Chapter 5. As the platform stiffness inclined, the stability limits of sedentary middle-aged adults decreased, and among all lower limb muscle groups, only the activity of rectus femoris increased. There is left-right asymmetry of muscle response found in the biceps femoris and medial gastrocnemius. The data was limited to sedentary lifestyle middle-aged adults. The study protocol used barefoot standing.

In addition, the changes to postural control and lower limb muscle activation among transtibial amputee during support surface perturbation were examined in Chapter 6. The inclination of support surface perturbation increased the center of pressure sway of directional stability during dynamic standing. Transtibial amputees generated higher muscle activation at their intact leg, where it only found in the biceps femoris. All the amputees lived a sedentary lifestyle. The data may not be generalized to experienced amputees with higher activity levels. Leg dominance was assumed to be negligible in this study.

Furthermore, Chapter 7 aimed to examine the possible balance-related factors associated with postural balance of transtibial amputee during stance posture. The balance control for TTA was negatively associated with the inclination of body mass index during static stance. There is a tendency of overweight amputees to exhibit greater postural sway than normal weight amputees in dynamic stance. A significant relationship was found for body weight and BMI with the measurement site of mid-foot and heel for two-point discrimination sensitivity. No variables were found to be possible balance-related factors for TTAs. The data were not generalized for first-fitted prosthetic users. The confidence level for the transtibial amputees and healthy adults are identical. The type of prosthetic usage was not standardized.

Therefore, the main outcome has shown that physical factors such as BMI could impact on balance performance, but not the biomechanical constraints such as quiet standing posture and area of BoS among healthy adults. For the transtibial amputee, the findings are clearly revealed that most of the anthropometric variables do not influence the balance performance. Only the changes of tactile sensitivity level would impact on balance performance. The ability of postural control for transtibial amputees will not be associated with the physiological factors such as height, weight, vibration sensitivity, lower limb muscle activation and body fat percentage, and biomechanical factors such as standing posture and surface stiffness.

The work presented in this dissertation could be improved through the increase of sample size. A larger sample size of subjects would provide a higher statistical confidence in the findings. Based on the current findings, a series of rehabilitation programs could be developed. The current findings would provide insight into how transtibial amputees may efficiently improve their balance performance from the time before amputation. Such knowledge would be valuable in developing new training or rehabilitation program for improving balance control.

The specific problems relating to biomechanics in stance could be investigated. The stance problems such as overloading at the intact leg, inflammation of the residual stump, and changes in body alignment also may influence the stance control. Future work could investigate the main cause that results in the stance problem mentioned above. It will offer new insights into how such causes can be avoided and treated. The stance postural stability for other types of amputation also could be investigated in greater detail. The current findings may serve as baseline information for the future studies which investigate stance postural stability with other disability groups.

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Ku, P. X., Abu Osman, N. A., & Wan Abas, W. A. The directional stability and lower limb muscle activity during dynamic standing among transtibial amputees. (Submitted)

Ku, P. X., Abu Osman, N. A., & Wan Abas, W. A. The changes of balance control for transtibial amputee in static and dynamic standing. (Submitted)

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