INVESTIGATION OF THE USE OF THE ARMS IN RECOVERING FROM POSTURAL PERTURBATIONS

A THESIS SUBMITTED TO
THE GRADUATE SCHOOL OF SOCIAL SCIENCES
OF
MIDDLE EAST TECHNICAL UNIVERSITY

BY

EMRE AK

IN PARTIAL FULFILLMENT OF THE REQUIREMENTS
FOR
THE DEGREE OF DOCTOR OF PHILOSOPHY
IN
THE DEPARTMENT OF PHYSICAL EDUCATION AND SPORT

JANUARY 2014

Approval of the Graduate School of Social Sciences

Prof. Dr. Meliha Altunışık
Director

I certify that this thesis satisfies all the requirements as a thesis for the degree of Doctor of Philosophy.

Prof. Dr. M. Settar Koçak
Head of Department

This is to certify that we have read this thesis and that in our opinion it is fully adequate, in scope and quality, as a thesis for the degree of Doctor of Philosophy.

Prof. Dr. Feza Korkusuz
Co-Supervisor

Prof. Dr. M. Settar Koçak
Supervisor

Examining Committee Members

Prof. Dr. Settar Koçak	(METU, PES)	
Assoc. Prof. Dr. Stephen J. Piazza	(PSU, USA)	
Assoc. Prof. Dr. Hayri Ertan	(AU, FSS)	
Assist. Prof. Dr. Sadettin Kirazcı	(METU, PES)	
Assist. Prof. Dr. Yeşim Çapa Aydın	(METU, EDS	

I hereby declare that all information in this document has been obtained and presented in accordance with academic rules and ethical conduct. I also declare that, as required by these rules and conduct, I have fully cited and referenced all material and results that are not original to this work.

Name, Last name : Emre Ak

Signature :

ABSTRACT

INVESTIGATION OF THE USE OF THE ARMS

IN RECOVERING FROM POSTURAL PERTURBATIONS

Ak, Emre

Ph.D., Department of Physical Education & Sports

Supervisor: Prof. Dr. M. Settar Kocak

Co-Supervisor: Prof. Dr. Feza Korkusuz

January 2014, 115 pages

Despite extensive preventive efforts, falls continue to be a major source of morbidity and mortality Understanding the methods used to recover from falling is important to develop necessary prevention techniques. Arm movements play an important role in recovering from postural perturbations.

This study aimed to understand the effects of arm rotations in three different levels of perturbation. The participants leaned forward in 5.5, 6.5 and 7.5 degrees and suddenly released by switching off the electromagnet attached on their back. A six camera motion analysis system and a force plate were used to record the kinetic and kinematic variables. Visual 3d software was used to create subject specific models. The angular momentum of the arms about the shoulder and MTP joints were calculated. The differences between 5.5 and 6.5 degrees were compared in terms of the angular momentum and the angular velocity of the arms. None of the participants were able to recover from 7.5 degrees.

Significant differences were found between 5.5 and 6.5 degrees in peak negative angular momentum of the arms, and negative and positive angular velocity of the forearms. There was no significant difference between the ground reaction forces, steady state times and shoulder torques. The relationships between angular velocity of the arms and Body Mass, Stature, Moments of Inertia of the forearms and BMI were found to be significant for 6.5 degrees. In conclusion, it can be speculated that central nervous system does some very complicated calculation based on the person's physical characteristics to create necessary kinetic and kinematic effects to recover from falling.

Key words: Fall Recovery, balance, angular momentum

ÖZ

DENGE KAYBI SIRASINDA DENGENIN YENIDEN KAZANILMASI ICIN KOLLARIN KULLANILMASININ ARASTIRILMASI

Ak, Emre

Doktora, Beden Eğitimi ve Spor Bölümü

Tez Danışmanı: Prof. Dr. M. Settar Koçak

Ortak Tez Danışmanı: Prof. Dr. Feza Korkusuz

Ocak 2014, 115 sayfa

Birçok koruyucu önlemler alınmasına rağmen düşmeler büyük oranda sakatlıklara ve hatta ölümlere sebep olan etkenlerin başında gelmektedir. Düşmeleri önlemek için uygulanan stratejilerin araştırılması, bu çok önemli sorunun oluş sıklığının azaltılması için yöntemlerin geliştirilmesine yardımcı olacaktır.

Bu çalışmada farklı seviyelerde denge kaybı sırasında yapılan kol rotasyonlarının dengenin kazanılmasına olan etkisileri incelenmiştir. Katılımcılar bellerine bağlı olan elektromagnet sayesinde 5.5, 6.5 ve 7.5 derecelik açılarda öne doğru eğildikten sonra beklemedikleri bir anda serbest bırakılmışlardır. Öne doğru düşme eğilimi gösterdikleri sırada dengelerini yeniden kazanmak amacıyla adım almadan önce istedikleri şekilde vücut uzuvlarını kullanarak dengelerini geri kazanmaya çalışmaları istenmiştir. Bu sırada yaptıkları hareketler altı kameralı hareket analizi sistemi ve kuvvet platformu ile kaydedilmiştir. Elde edilen 3 boyutlu marker pozisyonları kullanılarak herbir katılımcıya özel iskelet model oluşturulmuştur. Gerekli olan verilen kaydedildikten sonra MATLAB kodu kullanılarak analizler yapılmıştır. Hiçbir katılımcı 7.5 derecede yapılan denemelerde toparlanmayı başaramadıkları için bu açı analizler sırasında kullanılmamıştır.

Yapılan analizler sonucunda kolların vücudun ön tarafında aşağı doğru hareketi sırasında oluşan negatif açısal momentumlar arasında anlamlı farklar olduğu bulunmuştur. Ön kolların açısal hız değişimi ile vücut ağırlığı, beden kitle indeksi, boy uzunluğu ve ön kolların hareketsizlik momenti arasındaki ilişki incelendiğinde ise sadece 6.5 derecede anlamlı ilişki bulunmuştur. Bireyler denge kaybının şiddeti artıkça kolarını daha hızlı çevirirken, bireyin vücut ağırlıüı arttıkça kollarını çevirme hızının düştüğü bulunmuştur. Bu da denge kaybı gibi ani durumlar karşısında yapılan hareketlerin tamamen refleksif değil merkezi sinir sistemi tarafından planlı bir şekilde ortaya koyulduğunu gösterdiği düşünülmektedir.

Anahtar Kelimeler: Denge kaybı, toparlanma, açısal momentum

to GÜLŞAH & BADE

for their unconditional LOVE…

ACKNOWLEDGEMENTS

Many thanks to:

- My advisor, Prof. Dr. Steve Piazza, first accepting me as a visiting scholar to Penn State University, Biomechanics Laboratory and his guidance, advice and patience during my visit.
- My advisor and head of department Prof. Dr. M. Settar Koçak for his understanding and guidance throughout my PhD and Research Assistantship in the department.
- My ex-advisor and co-advisor Prof. Dr. Feza Korkusuz for his guidance throughout my PhD. I appreciate the doors he has opened for me.
- Emre Akçay and Uğur Ödek, Dicle Ekinci for their supports to make my thesis better.
- All my former research mates, Assist. Prof. Dr. Yaşar Salcı, Dr. Özgür Çelik and Dr. Ahmet Yıldırım for their understanding and ready to help in research and non-research matters.
- My lovely daughter and wife for their patience and complimentary love.
- My family for always being there as a source of encouragement.

TABLE OF CONTENTS

LIST OF TABLES

TABLES

LIST OF FIGURES/ILLUSTRATIONS/SCHEMES

FIGURES

CHAPTER I

INTRODUCTION

1.1. Background of the Study

According to WHO (World Health Organization) falls are the second leading cause of accidental or unintentional injuries or deaths worldwide after road accidents (WHO, 2012). In Canada, the leading causes of injury among all major injury cases in the 2010–2011 were reported as unintentional falls, which accounted for 39 % of all cases (n = 5,948) (CIHI, 2013). It was also reported that 424.000 individuals die from falls globally and mostly from low and middle income countries. It is more surprising that 37.3 million falls are severe enough to require medical attention each year (WHO, 2012).

The financial costs from fall-related injuries are substantial. For people aged 65 years or older, the average health system cost per fall injury in the Republic of Finland and Australia are US$ 3,611 and US$ 1,049 respectively. SMARTRISK reported that the direct cost of falls is about $ 4,457 million, indirect cost is about $ 1,698 million, creating a total cost of $ 6,155 million in Canada (SMARTRISK, 2004). The implementation of effective prevention strategies can make approximately 20% reduction in the incidence of falls among children under 10 and this could create a net savings of over US$ 120 million each year (WHO, 2012).

Falls are prominent among the external causes of unintentional injury. They are coded as E880-E888 in International Classification of Disease-9 (ICD-9), and as W00-W19 in ICD-10, which include a wide range of falls, including those on the same level, upper level, and other unspecified falls (WHO, 2012). A fall is defined as an event which results in a person coming to rest inadvertently on the ground or floor or other level. To stay upright, the human body uses exquisitely complicated and delicate feedback control, operating through the sensory and central nervous system (Eurich & Milton, 1996). The muscles responsible for maintaining balance receive signals that convey information on how much force to apply through which muscles. This information is initially supplied to the central nervous system by neurons located in various joints and the soles of the feet (Moss & Milton, 2003).

The causes of falls include vestibular, somotosensory or visual problems, age related Central Nervous System (CNS) processing latencies, deterioration of neural pathways used for motor control, loss of muscoskeletal integrity and other problems (Horak, Shupert, & Mirka, 1989). It was also found that healthy older adults exhibit decreased coordination between postural reflexes and voluntary movement, which in turn contribute to falling in daily activities (Stelmach, Phillips, Difabio, & Teasdale, 1989). When subjected to involuntary perturbations, older adults have been observed to overactivate their muscles not necessarily to recover from falling and balance stabilization, while young adults use their muscles in a controlled manner just enough to recover from falling if possible (Manchester, Woollacott, Zederbauerhylton, & Marin, 1989).

Among all types of falls, slipping and tripping are the most common causes, accounting for 30 to 50 percent of all falls, (Cumming & Klineberg, 1994; Gabell, Simons, & Nayak, 1985; Topper, Maki, & Holliday, 1993). Falls occur as a result of interaction of different risk factors. Those are categorized into four dimensions: biological, behavioral, environmental and socioeconomic factors. Biological factors embrace characteristics of individuals that are pertaining to the body, such as; age, gender and race, which are non-modifiable biological factors. Behavioral risk factors include those concerning human actions, emotions or daily choices. They are potentially modifiable. For example, risky behavior such as the intake of multiple medications, excess alcohol use, and sedentary behavior can be modified through strategic interventions for behavioral change. Environmental factors encapsulate the interplay of individuals' physical conditions and the surrounding environment, including home hazards and hazardous features in public environment. These factors are not by themselves cause of falls – rather, the interaction between other factors and their exposure to environmental ones are the reason for most of the falls. Home risk factors include narrow steps, slippery surfaces of stairs, looser rugs and insufficient lighting. Poor building design, slippery floor, cracked or uneven sidewalks, and poor lightening in public places are such hazards to injurious falls. Socioeconomic risk factors are those related to influence of social conditions and economic status of individuals as well as the capacity of the community to challenge them. These factors include: low income, low education, inadequate housing, lack of social interaction, limited access to health and social care especially in remote areas, and lack of community resources. The interaction of biological factors with behavioral and environmental risks increases the risk of falling. For example, the loss of muscle strength leads to a loss of function and to a higher level of frailty, which intensifies the risk of falling due to some environmental hazards (WHO, 2012).

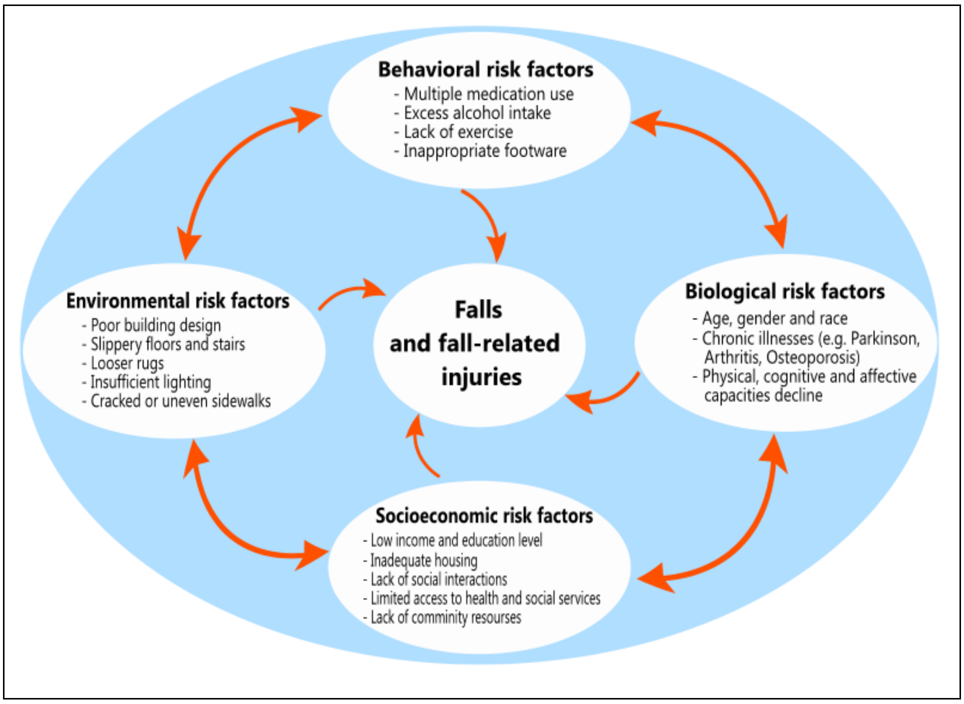


Figure 1. Risk Factor model for falls.

Different types of experimental procedures have been used to understand the strategies used to recover from a fall. These paradigms include the introduction of an unexpected obstacle during walking (Pavol, Owings, Foley, & Grabiner, 1999; M. Pijnappels, Bobbert, & van Dieen, 2004), slippery surface (Cham & Redfern, 2001; Troy & Grabiner, 2006), or constraint of swing leg via a tether attached at the ankle (Forner Cordero, Koopman, & van der Helm, 2003; Smeesters, Hayes, & McMahon, 2001). During upright stance, a sudden pull at the waist (Luchies, Alexander, Schultz, & Ashton-Miller, 1994; Rogers, Hedman, Johnson, Cain, & Hanke, 2001) or a sudden translation of the floor (McIlroy & Maki, 1996; Pavol, Runtz, Edwards, & Pai, 2002) have been employed (Hsiao-Wecksler, 2008).

To be able to stay in upright and keep body in balance, the CNS uses sensory information from vision. Vestibular, and somotosensory inputs produce appropriate motor commands to keep the body's Center of Mass (CoM) within the Base of Support (BoS) (Dawn Crystal Mackey, 2004). The strategies used for recovering from a fall are categorized into two main titles: feet-in-place (FIP) and change-in-support (CIS) reactions. Feet-in-place strategy is characterized by an unchanging base of support. Two main feet-in-place strategy have been identified for antero-posterior (AP) perturbations. The ankle and the hip muscles are used to stabilize the CoM within the base of support. Arms are also used to keep CoM within the BoS. After a postural perturbation, the ankle muscles (gastrocnameus and tibialis antreior) are activated, followed by the activation of proximal muscles on the hip. The ankle strategy is termed as 'automatic' postural response. The second strategy is change-in-support strategy which basically changes the base of support area. This includes stepping or grasping. If two strategies are to be compared, the ankle strategy is used for small-magnitude perturbations while change-in-support strategy is used for large-magnitude of perturbations. When no instruction is given, people prefer change-in-support strategy rather than using hip strategy.

All studies mentioned above have focused on feet-in-place reactions including ankle and hip strategies, change-in-support reactions, such as single step, multiple steps or grasping. However, there is limited information regarding the use of arms in fall recovery and whether arms are modulated depending on the severity of perturbation. Arm reactions following a forward fall was the focus of this work.

The question here is whether the central nervous system modulates the body parts to take necessary actions or it is a reflexive movement that maximal effort is being used.

Cerebellum is the main part in the central nervous system that is responsible for the balance control. The cerebellum is a part of the central nervous system with a rich neural network that regulates movement control by influencing timing in motor activities and smooth and rapid progression between movements. The cerebellum has neural connections with other parts of the brain and the peripheral parts of the body and it continuously receives sensory information from the bones, joints and muscles about their position, rate and direction of movement and forces acting on them. The cerebellum in turn conveys this information to the motor control centers of the cortex (motor cortex) setting the background tone and posture so that the cortex can execute new movements depending on the aim of the person. So, it is the cerebellum that gives detailed information to the motor cortex about the position and action of a limb and so the motor cortex can plan the next move. The cerebellum behaves like a computer between the motor cortex and the body parts. If there is any difference between these two parts, the cerebellum corrects the movement of the body part to control the action. So, movements although planned and executed by the motor cortex of the central nervous system, the cerebellum is responsible for regulating and smoothly controlling the movements. The cerebellum plays an important role in motor control and maintenance of balance in daily life and sports (Ananth, 2014).

Almost all sports require a high level of balance control. Many techniques require rapid arm or limb movements to keep the body in a balanced position. Since sports are full of unexpected events, the results from this study would give us important information understanding the mechanism in movement control especially in sudden actions. We control our movements in a very short and limited time, but the question here is whether we have a conscious processing in the Central Nervous System or it is an autonomic response or the combination. Therefore, this study would help us to understand the mechanisms under sudden and stressful conditions. Then, we can use the information and can develop different strategies both for recovering from balance perturbations and also for practicing balance for sporting actions.

1.2. Rationale of the Study

As stated above, falls are an important global issue, creating health and financial problems worldwide. There are many factors influencing the recovery from a fall. As people age, it becomes harder for them to recover from a fall. The impact of fall increases as people get older which may even lead to death.

There has been some research about falls that used different types of perturbations. These include, falls from height, using obstacles on the walkway, tripping, waist pulls and pushes, and tether release methods. In these studies, people tried to recover from falling by using their muscles, grasping an object around or taking single or multiple steps. Researchers have tried to understand the muscular, kinetic and also kinematic responses in these studies. However, no study has focused on the effects of arm movements on fall recovery. In this study, we focused on the arm movements and their contribution to fall recovery. Arm movements also have great influence on sport actions such as somersaulting, twisting, rotating, jumping and landing. A gymnast does repetitive rotations and then lands on the floor and suddenly stops. It's very interesting to see these athletes having such an angular velocity and angular momentum, controlling their body and suddenly stopping without losing balance. Similarly, ballerinas or figure skaters increase the speed of their rotation by decreasing the base of support and squeezing themselves to increase the speed of rotation, and then they open their arms and slow down. Volleyball players jump for a spike, hit the ball flexing their upper body and arms with a high speed and strength; and at the same time, flex their lower body to equalize the angular momentum so that they do not rotate in the air. This way, they land on without getting injured. The angular momentum applies for most of the sports and it's important to understand how the body modulates body parts for different conditions. So, this study will be a step to understand how the body modulates arms during a fall recovery to be able to apply the same principles to sports and understand the underlying mechanisms.

1.3. Research Question

Does the severity of the perturbation affect;

- the peak negative and positive angular momentums of the arms during fall recovery?
- the peak negative and positive angular velocity of the arms during fall recovery?
- the peak ground reaction force during fall recovery?
- the steady state times during fall recovery?
- the shoulder torques during fall recovery?

Does the forearm angular velocity have any relationship with the body mass, height, BMI, forearm moments of inertia during fall recovery?

1.4. Purpose of the Study

The purpose of this study was to investigate the effects of the modulation of the arms in fall recovery using tether release method in healthy collegiate students. This thesis addresses unexplored aspects of balance recovery strategies using the arms by using tether release method, which has received little attention when compared to the balance recovery strategies that involve, single or multiple steps, grasping an object or both. In particular, I sought to examine the influence of neuromuscular and behavioral variables on movement strategies to maintain balance and protective responses to avoid injury in the event of a fall.

1.5. Research Hypotheses

There will be differences between different lean angles in terms of the;

- peak angular momentum of the arms,
- peak positive and negative angular velocity of the forearms,

- peak ground reaction force
- steady state times and
- peak shoulder torques

There will be correlation between the peak angular velocity and the body mass, weight, BMI and the forearm moment of inertia's.

1.6. Delimitations

1- Participants consisted of 21-27 years old graduate students from Pennsylvania State University

2- The experimental protocol was the same for all participants.

3- All measurements were performed using the same set-up throughout the course of testing the participants.

4- This study demonstrates a forward fall by using a tether release method.

1.7. Limitations

1- Participants were not selected randomly.

2- Participants were limited to the graduate students in Pennsylvania State University.

3- Life history and the exercise activities of the participants were not taken into consideration.

4- The study was limited to the tether release perturbation.

1.8. Assumptions

1- The participants gave their best effort to recover from falling with their upper body before taking a step.

2- The forward lean angle was put in 5.5, 6.5 and 7.5 degrees.

3- The participants leaned forward by the means of tether and fully depended on the tether.

4- The release time of the tether was random.

1.9. Definition and Abbreviation of Terms

The following are definitions of the terms that were operationally defined throughout this study

WHO	: World Health Organization
ICD-9	: International Classification of Disease-9
CNS	: Central Nervous System
AP	: Antero-posterior
CoM	: Center of Mass
BoS	: Base of Support
FIP	: Feet-in-place
CIS	: Change-in-support
CoG	: Center of Gravity
LoG	: Line of gravity
VOR	: Vestibulo-ocular reflex
DALYs	: Disability-adjusted life years
EMG	: Electromyography
H_a	: Angular momentum about axis a through the center of gravity,
Σ	: Summation symbol,
I_i	: Moment of inertia of segments i about its own center of gravity,
ω_i	: Angular velocity of segment i,
m_i	: Mass of segment i,

$r_{i/cg}$: Distance from center of gravity of segment i to center of gravity of entire body,

$\omega_{i/cg}$: Angular velocity of ri/cg about the center of gravity of entire body.

rad/s : Radians per second

^{o}/s : Degrees per second

rpm : Revolutions per second ()

$\bar{\omega}$: Average angular velocity

$\Delta\theta$: Angular displacement

Δt : Time

θ_f : Final angular position, and

θ_i : Initial angular position.

L : Linear momentum,

m : Mass

v : Instantaneous velocity.

CHAPTER II

REVIEW OF LITERATURE

This chapter gives detailed information about the fall related studies. These are background information requires for understanding stability, balance, center of gravity and center of mass, base of support, control of posture, mechanics of balance, falls, epidemiology of falls, cause and risk factors of falls, balance recovery strategies, muscle activations used to recover balance, disturbances used to recover balance, whole body angular momentum, moments of inertia, angular velocity, and conservation of angular momentum.

2.1. Stability

Stability is the capacity of an object to return to equilibrium or to its original position after it has been displaced (M.McGinnis, 2013). Stability is also defined as the quality relating to the degree to which a body resists being upset or moved. The major factors that affect a person's stability are:

a. the area of the base of support

b. the relation of the line of gravity to the edge of the base

c. the height of the center of gravity and

d. the mass of the person("Stability and Balance," 2008).

In many sports, athletes do not want to be moved from a particular stance or position. The stability of a human body is affected by the height of the center of gravity, size of the base of support and the weight of the object. The base of support is the area within the lines connecting the outer perimeter of each of the points of support. There are two examples for base of support below (figure 2a and 2b). In the first picture, a ballerina stands in the tiptoe position. It can be seen that the base of support is limited with the area covered with the fingers, which is very small. On the other hand, a tennis player stands on both feet and the distance between her feet is about shoulder width, which makes the base of support bigger. It can be said that the stability of the tennis player is bigger than the ballerina, in other words, the possibility of the ballerina to lose her balance is higher than the tennis player. But of course with practice, they can perform better in time and have a perfect stability.

The stability can be controlled by changing the stance and body position. How do we initiate a walking step? The foot is not only lifted and placed in front of the body. We lean forward until the line of gravity falls in front of the feet and we lose our stability. We begin to fall forward, and we step with one foot to catch the fall and re-establish our stability. So, walking can be described as a series of falls and catches.

2.2. Balance

Balance is a physical ability that may be improved through purposeful practice. There are two types of balance:

a. Static balance, when a person remains over a relatively fixed base and

b. Dynamic balance, when a performer is in motion.

Minimal postural sway is the key factor in maintaining balance. A small amount of postural sway is inevitable but large postural sway might be an indication of decreased sensorimotor control.

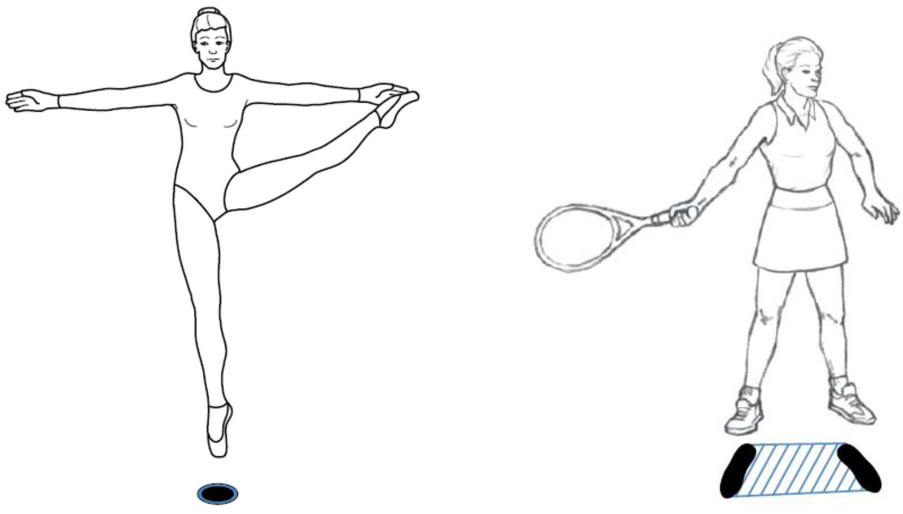
Maintaining balance requires coordination of vestibular, somatosensory, and visual systems. These systems were explained in the control of posture session.

2.3. Center of Gravity and Center of Mass

The CoM is practically the same as Center of Gravity (CoG). The center of gravity is defined in biomechanics as the imaginary point representing the weight center of an object, where all parts exactly balance each other. In the standing position, the body's CoG is located anterior to the second sacral vertebra. The CoG is closely related to body stability and balance. The line of gravity (LoG) is an imaginary line that extends vertically through the CoG to the center of the Earth. The interrelationship of the CoG and the LoG to the BoS determines the degree of stability of the body. On each body segment, the force of gravity acts according to the body segment mass (kg), and each segment has its own CoG (Arus, 2012).

2.4. *Base of Support*

The BoS represents the area of body part(s) in contact with a resistive surface that provides a reaction force to the applied force of the body. In other words, the BoS of the body is the area occupied under the body (e.g., in standing position) and describes one continuous line united with the outer edge of the body's contact with the ground. The BoS is extremely important in any sport and especially in martial arts (Arus, 2012) (figure 2).



a b

Figure 2. Base of Support for Ballerina and Tennis Player

2.5. *Control of Posture*

Sense of balance (Equilibrioception) is one of the physiological senses. It is important for both humans and animals to avoid falling over during walking or standing still.

A number of body systems works together for the balance: the eyes (visual system), ears (vestibular system) (figure 3) and the body's sense of location in space (proprioception) ideally need to be intact. The vestibular system, the region of the inner ear where three semicircular canals converge, works with the visual system to keep objects in focus when the head is moving. This is called the vestibulo-ocular reflex (VOR). The balance system works with the visual and skeletal systems (the muscles, joints and their sensors) to maintain orientation or balance. Visual signals sent to the brain about the body's position in relation to its surroundings are processed by the brain and compared with the information coming from the vestibular, visual and skeletal systems.

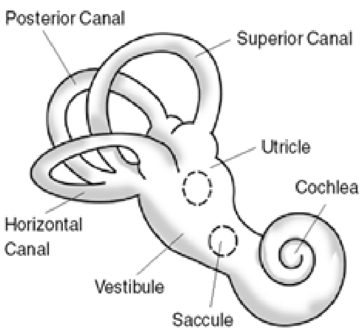


Figure 3. Vestibular System

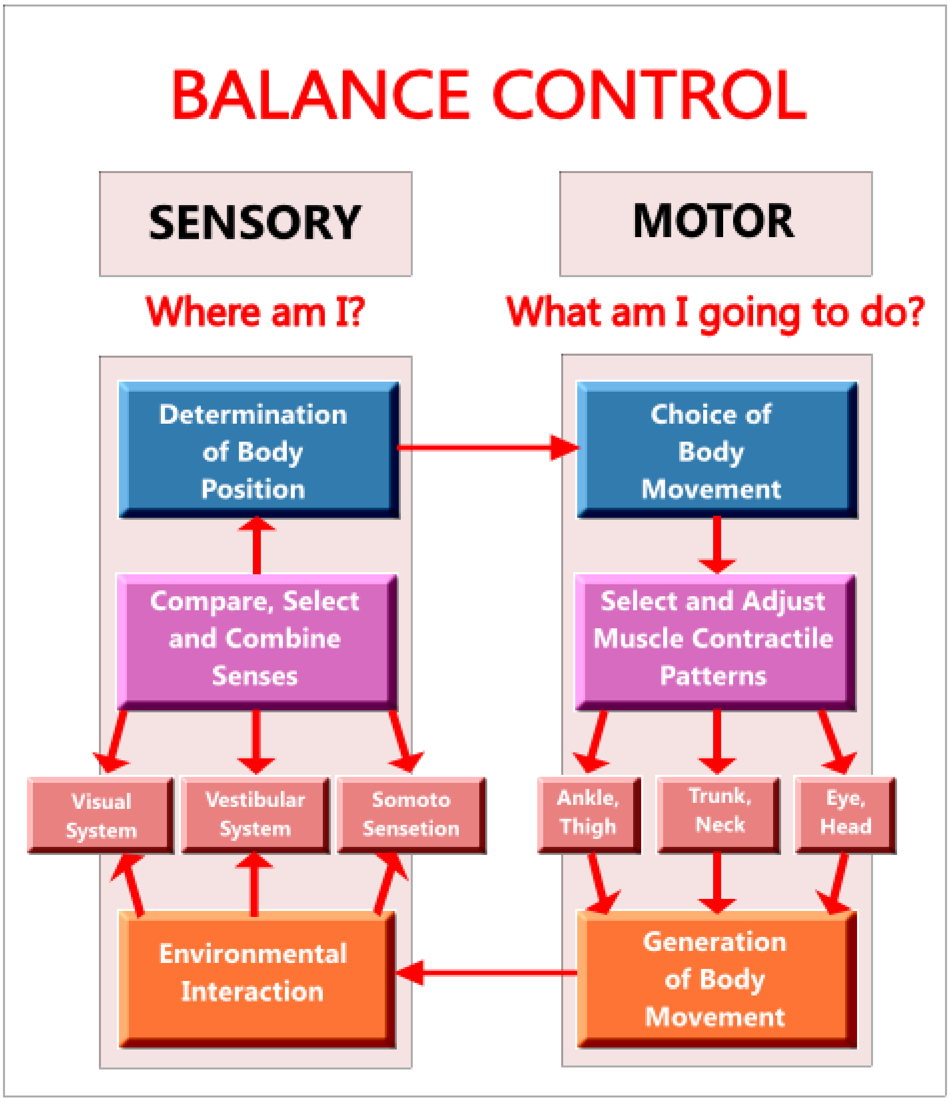


Figure 4. Balance Control System

Different forms of perturbation that disrupt a balanced body posture can be seen. The perturbations can be in different directions or magnitudes. Perturbations are not only slips, trips or bumping into some objects around us. It can be also be a very small volitional movement during a dynamic activity (Maki & McIlroy, 1997). Therefore, central nervous system is the one to integrate the incoming sensory information and then rapidly initiate and modulate the most appropriate balance correction to avoid falls and control the posture (Carpenter, Allum, & Honegger, 1999).

In any type of perturbation, sensory information must be integrated from a variety of sources including somotosensory, visual and vestibular pathways to make appropriate balance corrections. It is not always necessary to get information from all sensory pathways. When a perturbation occurs, information from any of the sensory pathway may help to trigger the appropriate response (Horak, Shumway-Cook, Crowe, & Black, 1988) (figure 5). This may suggest that the way sensory information is integrated for postural control must be flexible if there are changes in the environment (McCollum, Shupert, & Nashner, 1996).

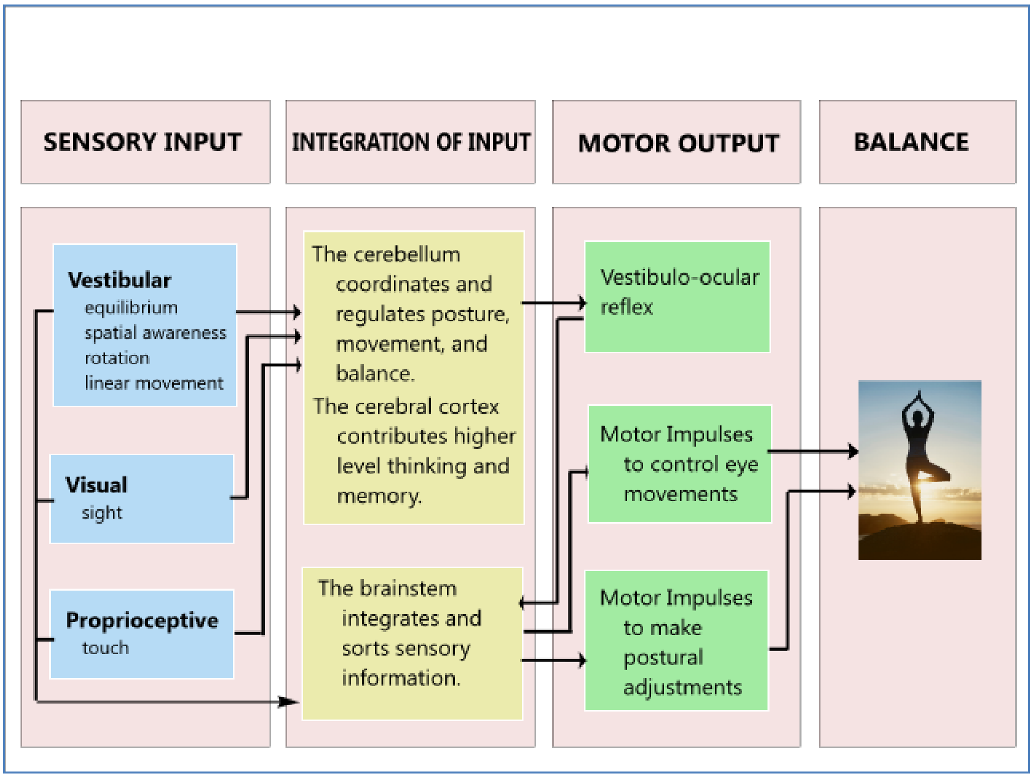


Figure 5. Balance Control System 2

2.6. *Mechanics of Balance*

The mechanics of the fall recovery can be divided into four categories (Hayes et al., 1996). The first phase is the (1) *balance maintenance stage* in which the person holds the centre of gravity within its base of support. Second phase is the (2) *initiation stage*, which involves loss-of-balance and potential attempts to regain upright posture; (3) a descent stage that involves preparations for landing or contact; (4) a contact stage where impact occurs between the body parts and the ground which can be a single step. This involves holding somewhere or an unsuccessful recovery resulting in the generation of reaction forces and absorption and/or dissipation of the body's kinetic energy.

2.7. Falls

A fall is defined as an event which results in a person coming to rest inadvertently on the ground or floor or other lower level (WHO, 2012). Fall-related injuries may be fatal or non-fatal though most are non-fatal. For example, of children in the People's Republic of China, for every death due to a fall, there are 4 cases of permanent disability, 13 cases requiring hospitalization for more than 10 days, 24 cases requiring hospitalization for 1–9 days and 690 cases seeking medical care or absenteeism from work/school.

Several studies demonstrated that age is one of the key risk factors for falls due to physical, sensory, and cognitive changes associated with ageing, in combination with environments that are not adapted for an aging population (M. Pijnappels et al., 2004; Mirjam Pijnappels, Kingma, Wezenberg, Reurink, & Dieën, 2009). Older people have the highest risk of death or serious injury arising from a fall and the risk increases with age. For example, in the United States of America, 20–30% of older people who fall suffer moderate to severe injuries such as bruises, hip fractures, or head traumas.

Another high risk group is children. Childhood falls occur largely as a result of their evolving developmental stages, innate curiosity of their surroundings, and increasing levels of independence that coincide with more challenging behaviors commonly referred to as ‘risk taking’. While inadequate adult supervision is a commonly cited risk factor, the circumstances are often complex, interacting with poverty, sole parenthood, and particularly hazardous environments.

Across all age groups and regions, both genders are at risk of falls. In some countries, it has been noted that males are more likely to die from a fall, while females suffer more non-fatal falls. Older women and younger children are especially prone to falls and increased injury severity. Worldwide, males consistently sustain higher death rates and DALYs lost. Possible explanations of the greater burden seen among males may include higher levels of risk-taking behaviors and hazards within occupations.

Most falls have no apparent link to environmental hazards (Morfitt, 1983), failed attempts during daily activities such as walking, running, raising and bending are common situations leading to falls (Nevitt & Cummings, 1994; Parker, Twemlow, & Pryor, 1996). Trips and slips are most common self-reported causes but people also reported “loss of balance”, “leg gave away”, “changed posture” as the situations they think is the reason of their falls (Blake et al., 1988; Cumming & Klineberg, 1994).

Other risk factors include:

- occupations at elevated heights or other hazardous working conditions;
- alcohol or substance use;
- socioeconomic factors including poverty, overcrowded housing, sole parenthood, young maternal age;
- underlying medical conditions, such as neurological, cardiac or other disabling conditions;
- side effects of medication, physical inactivity and loss of balance, particularly among older people;
- poor mobility, cognition, and vision, particularly among those living in an institution, such as a nursing home or chronic care facility;
- unsafe environments, particularly for those with poor balance and limited vision.

Fall prevention strategies should be comprehensive and multifaceted. They should prioritize research and public health initiatives to further define the burden, explore variable risk factors and utilize effective prevention strategies. They should support policies that create safer environments and reduce risk factors. They should promote engineering to remove the potential for falls, the training of health care providers on evidence-based prevention strategies; and the education of individuals and communities to build risk awareness.

Effective fall prevention programs aim to reduce the number of people who fall, the rate of falls and the severity of injury should a fall occur. For older individuals, fall prevention programs can include a number of components to identify and modify risk, such as:

- screening within living environments for risks for falls;
- clinical interventions to identify risk factors, such as medication review and modification, treatment of low blood pressure, Vitamin D and calcium supplementation, treatment of correctable visual impairment;
- home assessment and environmental modification for those with known risk factors or a history of falling;
- prescription of appropriate assistive devices to address physical and sensory impairments;
- muscle strengthening and balance retraining prescribed by a trained health professional;
- community-based group programs which may incorporate fall prevention education and Tai Chi-type exercises or dynamic balance and strength training;
- use of hip protectors for those at risk of a hip fracture due to a fall.

For children, effective interventions include multifaceted community programs; engineering modifications of nursery furniture, playground equipment, and other products; and legislation for the use of window guards. Other promising prevention strategies include: use of guard rails/gates, home visitation programs, mass public education campaigns, and training of individuals and communities in appropriate acute pediatric medical care should a fall occur (WHO, 2012).

2.8. Balance Recovery Strategies

As mentioned above, the incident of falling is one of the high risk factors of deaths and injuries. Therefore, being able to recover from a perturbation is important in avoiding any severe injuries in both daily life and sport activities. Most of the injuries occur because of a trip or slip and unable to take necessary recovery actions. If done effectively, a fall can be avoided or at least the severity of injury can be decreased.

In this section, five different balance recovery strategies have been described. These are; (a) ankle strategy, (b) hip strategy, (c) stepping or stumbling strategy, (d) grasping strategy, and (e) arm movement strategy. Strategies “a, b and e” strategies are considered as “feet in place” or “fixed support” strategies as they involve regulating movement of the body’s CoM with respect to a fixed base of support. The other two strategies are termed “change in support” strategy as the body’s CoM is controlled through a change in base of support (Woollacott & Shumway-Cook, 2002).

2.8.1. Ankle Strategy

This strategy is used when the body’s CoM is still inside the base of support. With this type of perturbation, the balance is restored by contracting the ankle joint muscles. The rotation of the body about the ankle joint shifts the body’s center of gravity when using this technique. The data that can be obtained from this response include recovery angle and ankle torque.

2.8.2. Hip Strategy

This strategy is also used when the center of gravity is still in the base of support. Depending on the side of the perturbation the hip flexor or extensor muscles can be contracted. If the perturbation is in forward direction, the hip extensors are contracted. If the perturbation is in backward direction then the hip flexors are contracted to recover from falling to keep the center of gravity in the base of support.

2.8.3. Stepping or Stumbling Strategy

Step responses are used when the body’s center of gravity is outside of the base of support. When the balance is lost and there is no chance of recovery with the ankle hip and other strategy, people use one or more steps depending on the severity of the perturbation to be able to stabilize the trunk. The direction of the perturbation determines the type of the step, whether forward, backward or sideways. COP, temporal, kinetic, and kinematic, and Electromyography (EMG) variables can be used with this type of studies. This strategy is the most commonly used one in sport activities.

2.8.4. Grasping Strategy

If possible, grasping is the first choice of people who lose the balance to be able to avoid a fall. It can be a person or any kind of stable object around. It is also called fixed support balancing reaction which provides early defense in loss of balance. It is very important to move the limbs so as to alter the base of support. It can be taking a step or reaching and grasping an object for support.

2.8.5. Arm Movement Strategy

Another foot in place technique to recover from balance is arm movements. Additional to ankle and hip muscle contractions, arms can be used to keep the center of gravity in the base of support. During a forward perturbation, the body creates a certain angular momentum. By reflex, the arms are raised or can be rotated (McIlroy & Maki, 1995). Arm movements also play an important role in creating more time to recover from falling by using the muscles and grasping an object around.

2.9. Muscle Activations used to Recover Balance

Muscle activation patterns during fall recovery have been investigated by various investigators (Nashner, 1976; Nashner, Woollacott, & Tuma, 1979). The ankle strategy aims to take the COG to a vertical, upright position by producing optimum torque about the ankle joint. Muscle activation begins with the ankle joint muscles and then thigh and trunk muscles are activated. This was shown by the onset latencies of gastrocnamius and soleus muscles, followed by the activation of hamstring and then paraspinal muscles (Murnaghan, 2008).

In hip strategy, people move their body as a two-linked pendulum. When the trunk moves anterior, the hip joint moves posterior to keep the COG within the base of support. The muscles are activated in proximal to distal sequence, while there is limited ankle muscle activation. Initially, abdominal and quadriceps muscles are activated, which makes the hip to flex, pushing the trunk forward and legs backward to stabilize the COG above the base of support (Horak & Nashner, 1986).

Most responses to recover from falls involve a mix of ankle and hip strategies (Runge, Shupert, Horak, & Zajac, 1999). These activation strategies depend on the velocity and the severity of the perturbation (Bothner & Jensen, 2001), the direction of the perturbation (Henry, Fung, & Horak, 1998) and the initial position of the person (Tokuno, Carpenter, Thorstensson, & Cresswell, 2006).

It was also found that tibialis anterior (TA) and medial gastrocnameus muscle are activated rapidly when the magnitude of perturbation is high. Similar results were found for ankle and arm muscles (McIlroy & Maki, 1995).

2.10. Disturbances Used for Balance Assessment

A number of different experimental paradigms have been used to simulate the loss of balance to examine the different responses. None of these methods are enough to make the perturbation as real as it is in daily life but they all aim to simulate the real life situations and make us understand how the body modulates its movements to keep it in balance again. The responses vary by taking a step, grasping an object or using ankle or hip strategies as mentioned above. These paradigms include walking of an unexpected obstacle (M. Pijnappels et al., 2004), slippery surface (Troy & Grabiner, 2006), constraint of the ankle of the swing leg (Forner Cordero et al., 2003), a sudden pull from the waist (Rogers et al., 2001), or sudden translation of the floor (Pavol et al., 2002). There is another method that was used in this study; tether release method is another way in which participants forward falls.

2.10.1. Platform Translations

This balance testing methodology (Maki, Holliday, & Fernie, 1987) was based on a posture control model which defined relative stability by the degree to which transient postural perturbation caused the COP on the feet to approach the limits of the base of support. The platform generated one degree of freedom in horizontal translation motion by the means of a motor and a ball screw using position and velocity feedback. The platform mechanisms were covered by a plywood base and safety handrails were mounted onto the undercarriage. Two force plates were placed side to side on the platform for each foot. Since the perturbations in daily life can occur in other directions they included a lateral component to the platform to simulate the lateral perturbation. The primary objective of their study was to determine how the postural control system resolves these potential interactions between swing-leg selection, preparatory unloading of the swing-leg and the swing trajectory formulation, in initiating compensatory stepping responses to lateral perturbation.

2.10.2. Pushes and Pulls on the Body

A waist-pulling device is used as a balance perturbation mechanism in experiments related to step responses and feet-in-place responses (Pai, Rogers, Patton, Cain, & Hanke, 1998). The apparatus consists of a belt worn around the waist, with a weight-dropping (or similar) mechanism delivering a disturbance through a cable attached to the waist belt. Waist pulling devices may incorporate pulley systems to direct balance disturbances in different directions (Hilliard et al., 2008; Mille et al., 2013). The waist pull device is commonly used to cause perturbations in the posterior or medio-lateral directions.

2.10.3. Tether Release Method

A Tether Release Method is also called lean and release disturbance and used to simulate the unbalanced body at the onset of a trip or slip. The person is held at an initial forward or backward lean angle. A cable is attached at about waist level and then suddenly released. Generally, an electromagnet is used to hold the participant at a certain angle and then the electromagnet is released and the participant tries to recover from falling by taking a step, grasping something around or using different strategies depending on the study. Mostly, a safety harness is worn to avoid falling but it does not support the person.

2.11. Moments of Inertia

Moment of inertia is the mass property of a rigid body that defines the torque needed for a desired change in angular velocity about an axis of rotation. Moment of inertia depends on the shape of the body and may be different around different axes of rotation. A larger moment of inertia around a given axis requires more torque to increase the rotation, or to stop the rotation, of a body about that axis. Moment of inertia depends on the amount and distribution of its mass, and can be found through the sum of moments of inertia of the masses making up the whole object, under the same conditions.

2.12. Angular Velocity

Angular velocity is defined as the rate of change in angular displacement. Its units of measurement are radians per second (rad/s), degrees per second (^{o}/s), revolutions per second (rpm), and so on. Angular velocity is abbreviated with Greek letter omega (ω). Angular velocity is a vector quantity, just like linear velocity, so it has direction associated with it. The direction of angular velocity is determined using right-hand thumb rule, as with angular displacement. Because angular velocity is a vector, a change in size of angular velocity or in the direction of its axis of rotation results in a change in the angular velocity.

Angular velocity is computed as the change in angular position (angular displacement) divided by the time. Mathematically,

$$\bar{\omega} = \frac{\Delta\theta}{\Delta t} = \frac{\theta_f - \theta_i}{\Delta t}$$

where

$\bar{\omega}$ = average angular velocity

$\Delta\theta$ = angular displacement

Δt = time,

θ_f = final angular position, and

θ_i = initial angular position.

Momentum

Linear momentum is the product of an object's mass and it's linear velocity. The faster the object moves, the more momentum it has. The larger a moving the object's mass, the more momentum it has.

$$L = mv$$

where

L = linear momentum,

m = mass, and

v = instantaneous velocity.

Newton's first law of motion basically states that the velocity of an object is constant if the net force acting on the object is zero. In sport and human motion, most objects we deal with have a constant mass. If the velocity of an object is constant then the angular momentum is constant since the mass does not change. Momentum is constant if the net external force is zero.

2.13. Whole Body Angular Momentum

During a human motion, some limbs rotate at different velocities and directions than others. How is the angular momentum determined? Mathematically, the angular momentum about an axis through the center of gravity of a multisegment object such as human body is defined by the equation of;

$$H_a = \sum(I_i\omega_i + m_i r^2_{i/cg}\omega_{i/cg})$$

where

H_a = angular momentum about axis a through the center of gravity,

Σ = summation symbol,

I_i = moment of inertia of segments i about its own center of gravity,

ω_i = angular velocity of segment i,

m_i = mass of segment i,

$r_{i/cg}$ = distance from center of gravity of segment i to center of gravity of entire body, and

$\omega_{i/cg}$ = angular velocity of ri/cg about the center of gravity of entire body.

The sum of angular momenta of all body segments gives an approximation of angular momentum of the entire body. When we examine the angular momentum during running, the left arm swings backwards and the right arm swings forward. The left leg swings forward and the right leg swings backward. Using the right-hand thumb rule, with the positive direction of the axis to the left, the angular momentum of the left arm is positive, the angular momentum of the right arm is negative, the angular momentum of the right leg is positive and the angular momentum of the left leg is negative. The trunk is not rotating, so its angular momentum is zero. If we use the equation above to approximate the total angular momentum of the body, it would appear that the angular momenta of the arms sum to zero (they cancel each other out) and the angular momenta of the legs sum to zero as well. The total angular momentum of the body is zero.

However, the body's moment of inertia is not always stable or the actions of symmetrical body parts do not always move in a total asymmetrical way. The moment of inertia is a variable and can be changed by altering the limb positions.

The body’s angular momentum also changes to accommodate the changes in the moment of inertia.

Conservation of angular momentum is well demonstrated by a figure skater doing a spin. The torque created by the friction between the ice and the skates is minimal and may be ignored. As the skater begins a spin, one leg and both arms may be held up and away from the body. The skater thus has a large moment of inertia about the longitudinal axis. As the spin progresses, though, the skater adducts the arms and legs, bringing them closer to the body, thus reducing the moment of inertia. Angular momentum remains the same, so the reduction in the moment of inertia must be accompanied by an increase in angular velocity, which is exactly what happens to the skater.

Gymnasts, figure skaters, dancers, divers, and other athletes use this principle of conservation of angular momentum to control their angular velocities when somersaulting and twisting. A gymnast tucks her body to speed rotation. A figure skater abducts his arms to slow down his spin. A dancer adducts her arms to speed up her spin. A diver extends from a pike to slow down somersaulting (M.McGinnis, 2013).

CHAPTER III

MATERIALS AND METHODS

This study was designed to investigate the effects of body movements on fall recover. The chapter contains the following parts; (1) Participants, (2) Instruments, (3) Experimental Set-up and Procedures, (4) Statistical Analysis.

3.1. Participants

Six males and four females from Pennsylvania State University volunteered to participate in this study (Age: 24.0 ± 2.0 years; Height: 1.73 ± 0.1 m; Body Mass: 74.8 ± 21.1 kg). After obtaining approval from Institutional Review Board (IRB) of Pennsylvania State University in State College, PA, USA, all participants gave informed consent (See Appendix A). All possible risks and consequences of the study were explained. Participants were screened and interviewed to ensure that they were in good general health, with no recent musculoskeletal injuries or sensory impairments. Two of the participants were excluded because of wrong initial lean angles. The experimental procedure was explained to the participants with all possible risks.

3.2. Instruments

All data collection took place at the Pennsylvania State University, The College of Health and Human Development, Department of Kinesiology, Biomechanics Laboratory. The facility contained two force plates embedded in the middle of the walkway which were used to measure the ground reaction force (model 9286AA, Kistler Instrument Corporation, Amherst, NY). Eight channels of analog output from each force plate were sampled at 1000 Hz. Eagle Digital System was used to collect 3 dimensional marker positions during the test.

3.2.1. *Eagle Digital System*

The system consists of six (6) Eagle Digital Cameras (figure 6) (Eagle Digital Real Time, Motion Analysis Corporation, Santa Rosa, CA), the EagleHub, and the EVaRT software (EVaRT 5.0, Motion Analysis Corporation, Santa Rosa, CA). Each Eagle Digital Camera has a resolution of 1.3 million pixels at 1280 x 1024 full resolutions. The resolution was 100 frames per second. Processing of the digital images is performed on the camera, rather than on a centralized computer system. The cameras connected to the Eagle Hub use fairly standard Ethernet wiring. Each camera communicates with the tracking computer where individual images are combined to create the 3D capture volume. The marker positions recorded by the software were post processed using the EVaRT software. The gaps between the time periods were filled and the time series were cropped to clear the unnecessary data. Then, the data were converted into c3d file format to be able to create subject specific models in Visual 3D software.

Figure 6. The Motion Analysis Corporation Eagle system

3.2.2. Kistler Force Plate

A Kistler force platform is used to acquire kinetic data (model 9286AA, Kistler Instrument Corporation, Amherst, NY). The dimensions of the force plate are 600x400mms. Four piezoelectric transducers located at the corners of the platform measure the applied forces. Forces are measured in vertical, anteroposterior (AP), and in the medio-lateral direction. The Kistler force platform is also capable of measuring the coordinates of the point at which the force is applied. This measure uses an x and y coordinates system.

3.2.3. Visual 3D Software

Visual 3D is advanced 3D analysis software available for biomechanical motion analysis (kinematics & kinetics). The 3 dimensional marker positions gathered from the Motion Analysis System are saved as c3d files and can be imported into the Visual 3D software. From the marker position, the 3D skeleton model of each participant can be created by using the necessary anthropometric measurements. The desired kinetic and kinematic calculations can be done via Visual 3D software. The pipeline helps us to create sequence of calculations to make it easy for each model. The data were exported as c3d file format which can be read by Visual 3D software. In visual 3D, subject specific models were created by using anthropometric measures taken from each participant. The process of model creation was shown with figures 7 to 13.

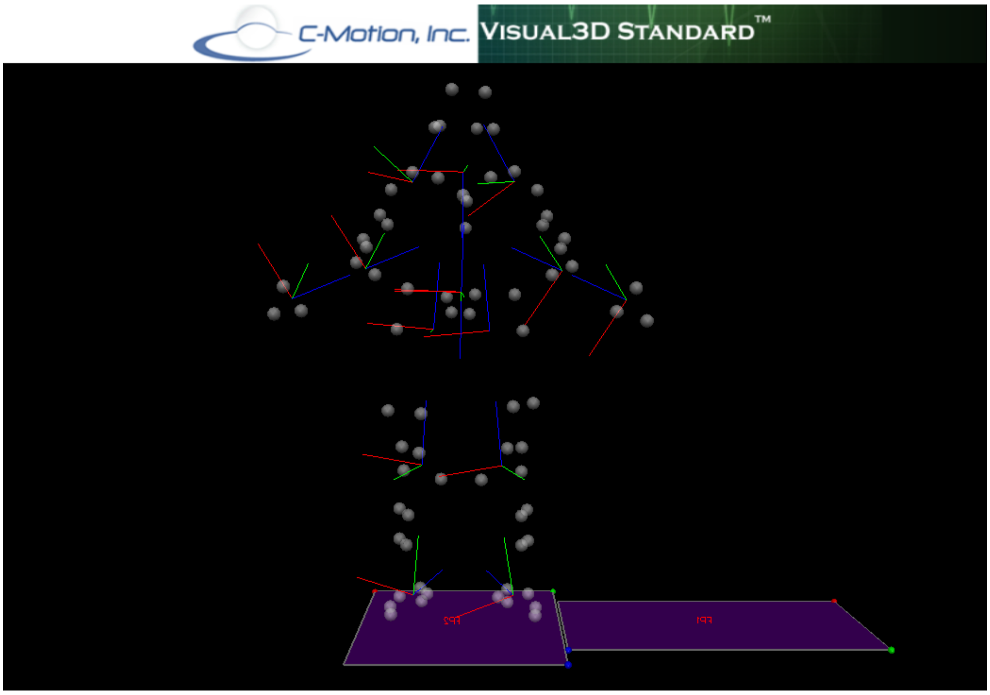


Figure 7. Marker Position View in Visual 3D

The participants were instructed to stand still for the static trial. This trial is used to create subject specific models in Visual3D software.

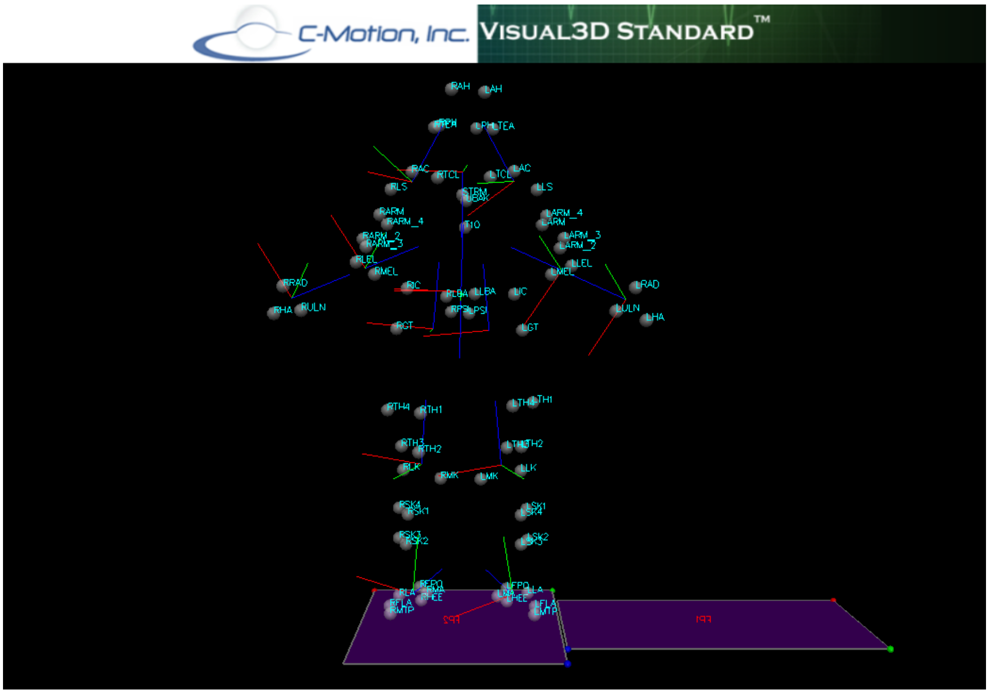


Figure 8. Marker ID's were given for each marker.

After recording the static trial in Motion Analyses System, the files were exported as c3d file formats to be used in Visual 3d. In Visual 3D, all the markers were given standard names that were recommended by the c-motion.

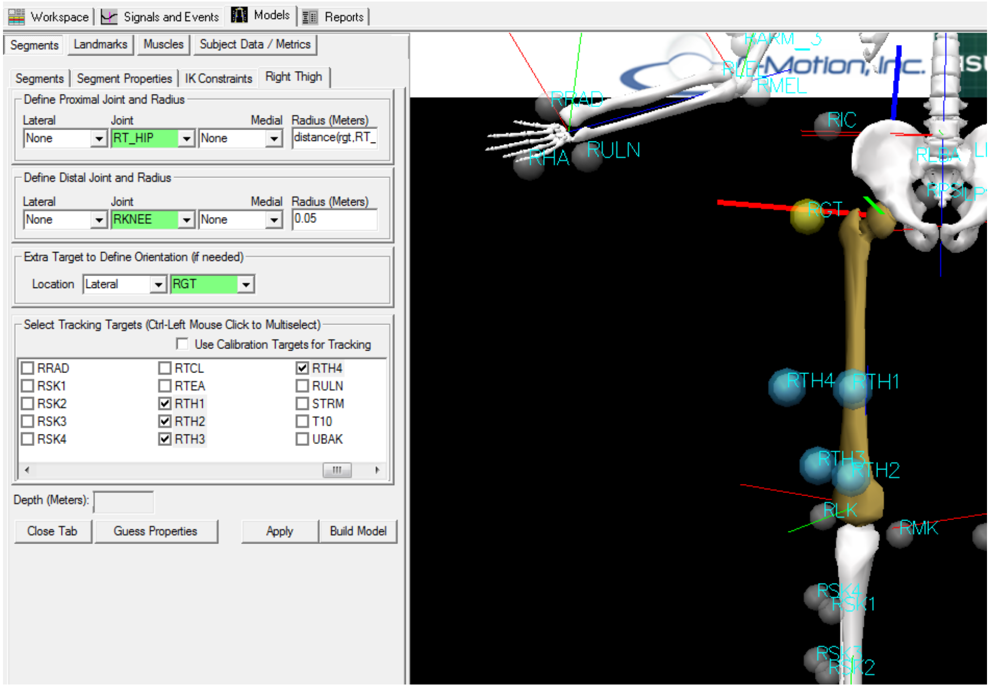


Figure 9. Segment Creation in Visual 3D

After naming each marker, the landmarks were created. They were used to create segments. Each segment was created using distal and proximal markers and tracking markers were assigned to each segment to track the movements of the segments. The tracking markers were put on the clusters to avoid movements with respect to each other.

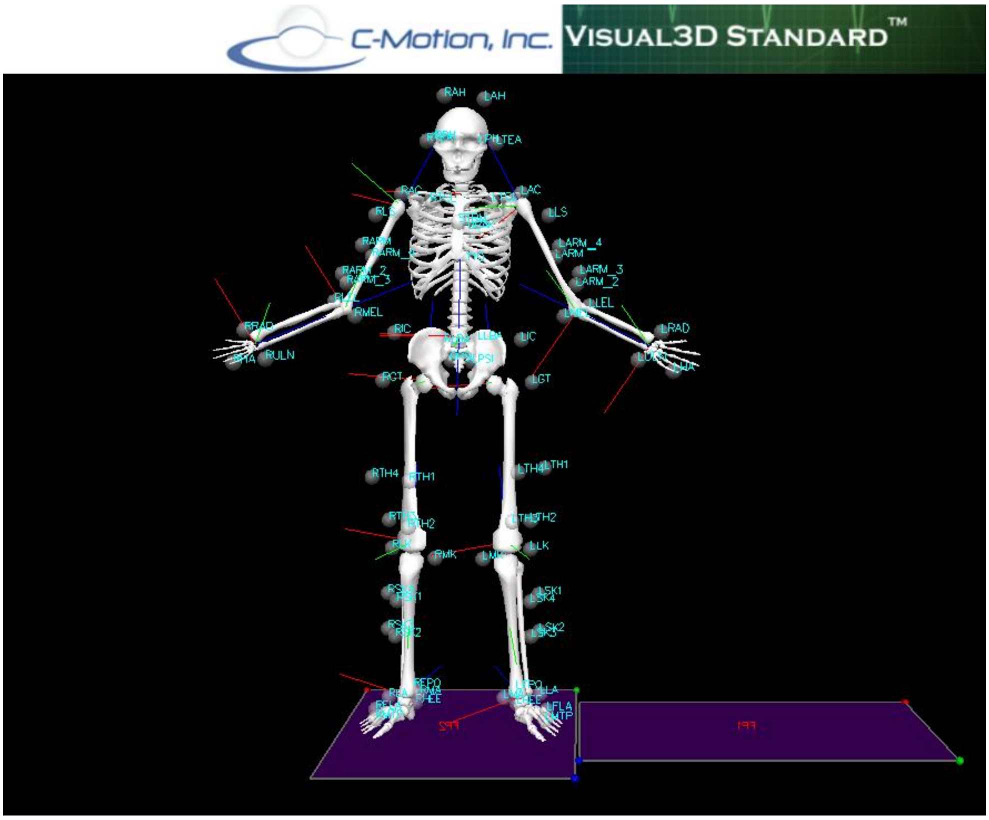


Figure 10. Full Body Model created in Visual 3D

The whole body skeletal model was created specific to each participant.

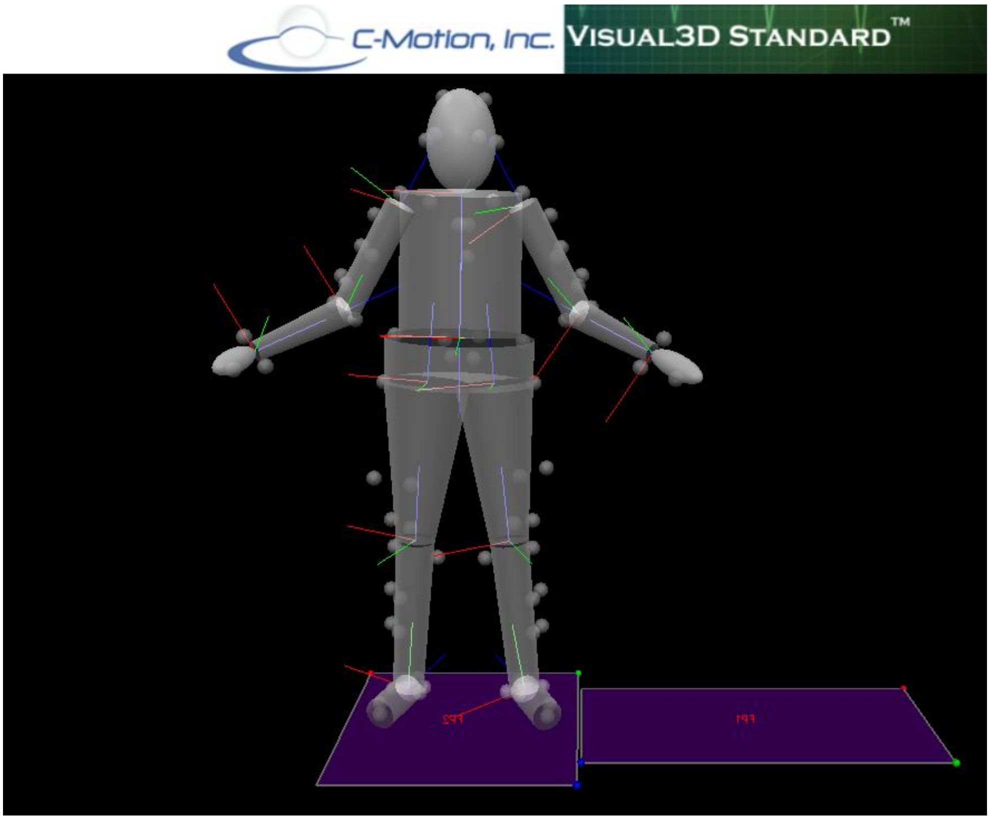


Figure 11. Segment Geometries created by using participant's body sizes.

When creating the skeletal model, the anthropometric measurements were also used to make the body size specific to each participant. This is important in calculating the segment inertias.

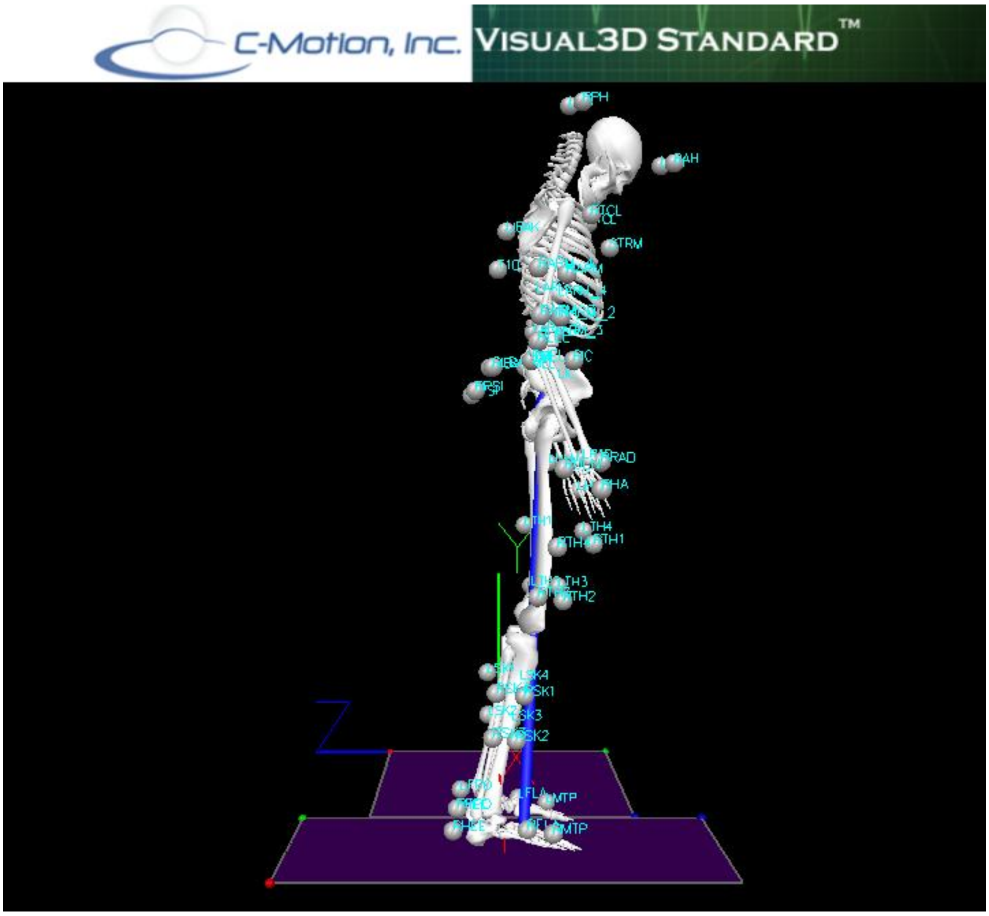


Figure 12. Test Protocol view in Visual 3D

After creating the subject specific models, the models were applied to each testing trial

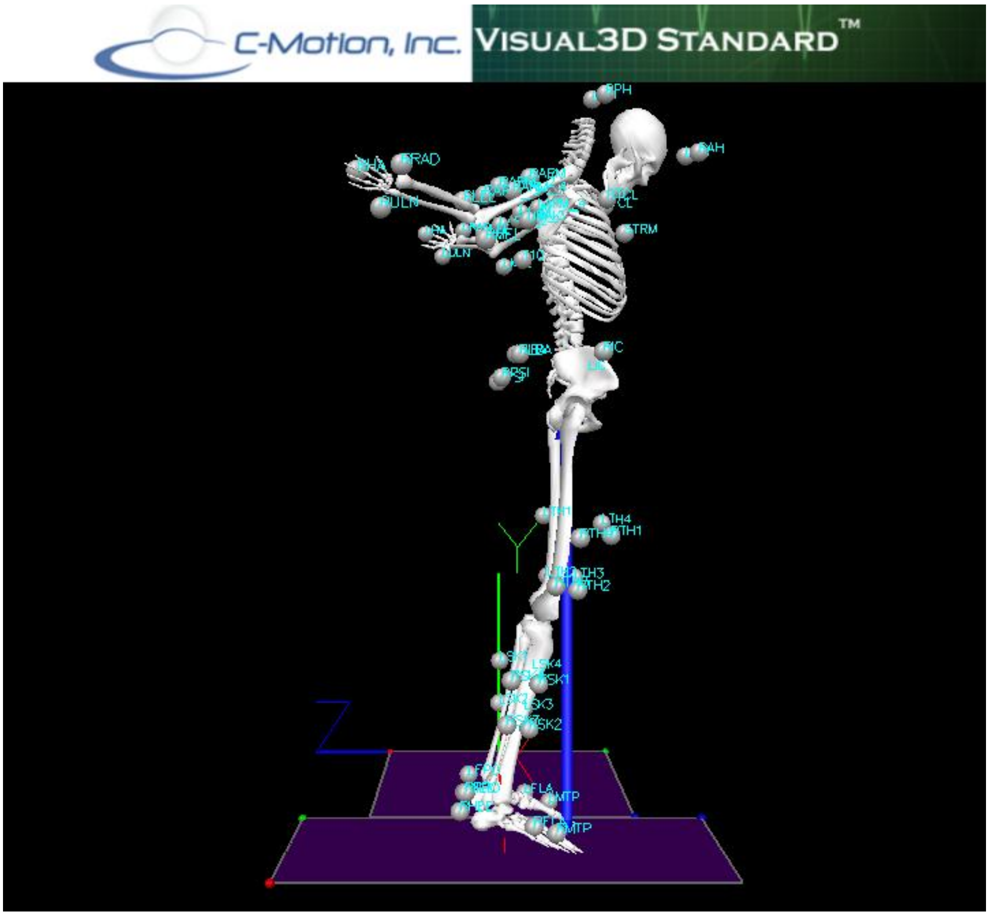


Figure 13. Participant trying to recover from falling: Arm Rotation

3.3. Experimental Setup and Procedures

The participants were asked to wear shorts or tights and short sleeve tops during the experimental protocol. They were barefoot during the trials. To avoid the possibility of subject fatigue, rest breaks of approximately 30 seconds were given between trials, and 3 minutes rest was given after each five trial.

Seventy two reflective markers were placed according to the conservative full body marker set, explained in Appendix C. This marker set was advised by the c-motion. After marker placement, the participants were asked to stand still on the force place facing forward with standard anatomical position. Standing trial was recorded using EVaRT software with 6 Eagle Cameras (Motion Analysis Inc.) and the force plate. Then, the markers that will be used only for segment creation but not for tracking the motion were removed. 54 markers were left on the participants.

The participants were initially held in a forward leaning position in 5.5, 6.5 and 7.5 degrees of angles by the means of a tether attached to the back of a pelvic belt supporting each subject while they kept their head, trunk and extremities aligned in the forward-leaning posture (figure 14). The magnitude of lean angle was calculated using simple trigonometry using the distance between the lateral malleolus of the right ankle and the wall, the height of the hook on the wall which the tether was attached, the distance between the leteral malleolus of the right ankle and the hook on the wall, the height of the hip joint where the other end of the tether was attached. The desired lean angle was put in the formula shown in the figure 9 and the desired length of the tether length was found. The lean angle was also checked by using the marker positions on the lateral malleolus, greater throcanter and acromion process after the experimental procedure. If there was any deviation more than 0.2 degrees, the trial was excluded. The subjects were instructed to keep their heels on the ground until the tether was released. A switch controlled electromagnet was used to hold the participants. These angles were selected because maximum lean angle that can be recovered by young people was reported as 7.2 degrees (D. C. Mackey & Robinovitch, 2006). None of the participants were able recover from 7.5 degrees. A safety harness was put on the participants, which was attached to the ceiling to avoid any fall during the trials. The participants were told to stay in a relaxed position with their body being straight.

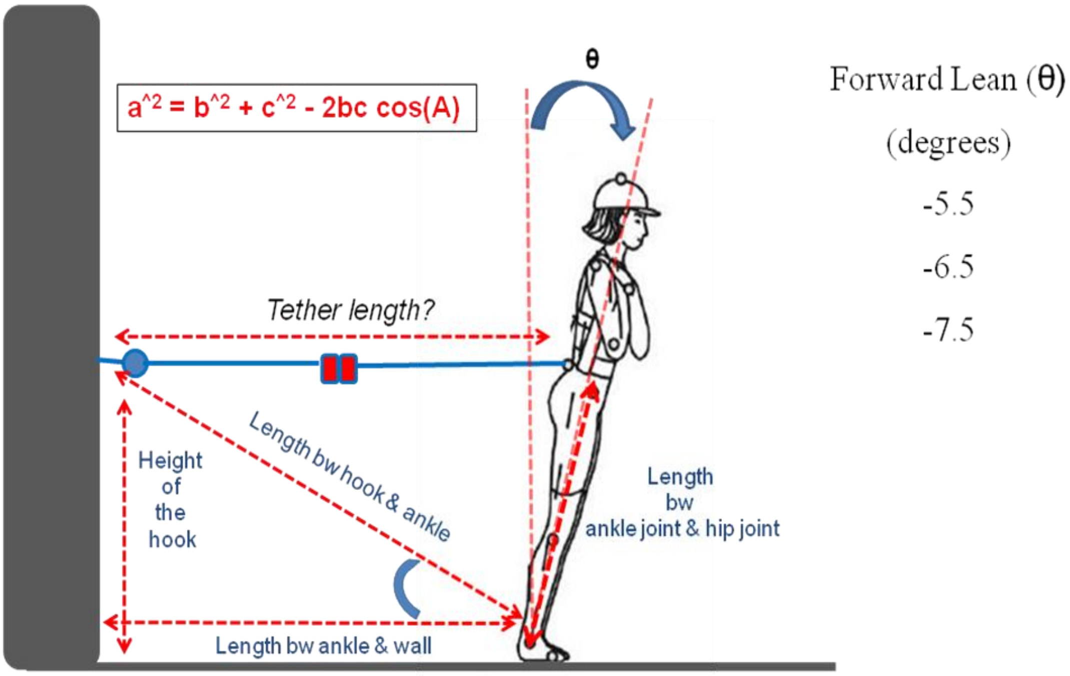


Figure 14. Lean Angle Calculation

Table 1. Experimental Procedure on Test Day

Steps	Duration
1- Explanation of the Procedure	5 min.
2- Sign the Informed Consent Form and Fill in the Information Sheet	3 min.
3- Placing the Markers	30 min
4- Finding the Length of the Tether for each Angle	10 min.
5- Recording the Trials and checking the data	20 min.
6- Removing the Markers	10 min.
7- Anthropometric Measurements	5 min.

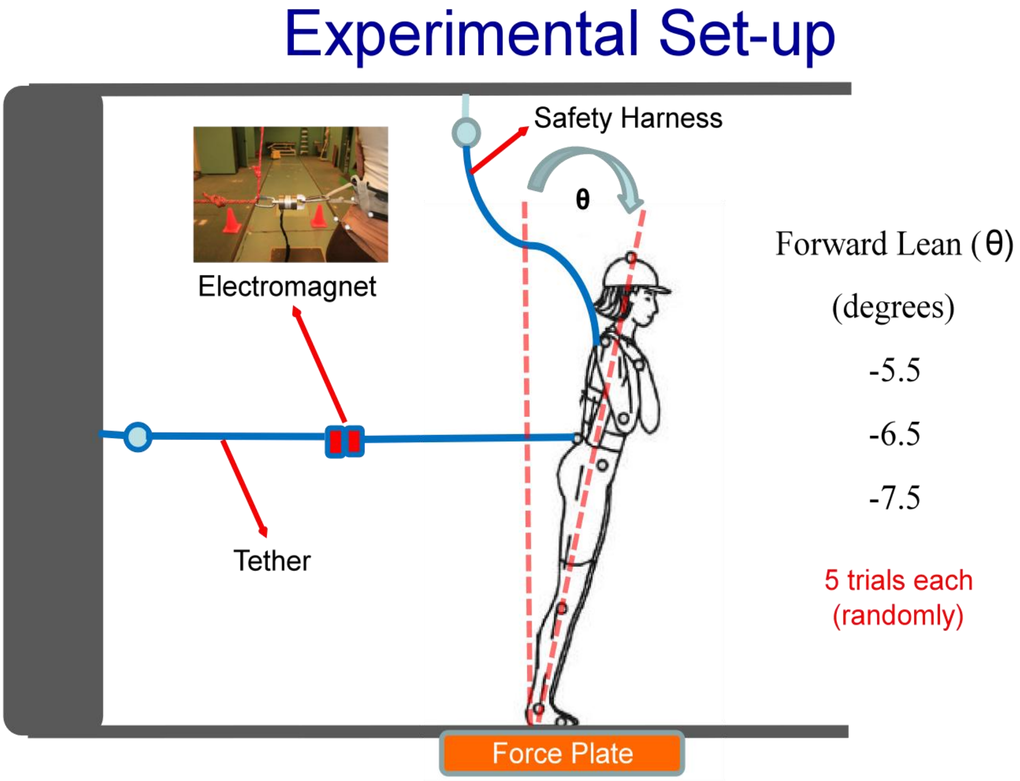


Figure 15. Experimental Set-up

Forward falls were induced by releasing electromagnet after a random time delay. Each angle (5.5, 6.5 and 7.5 degrees) was given five times randomly. The randomization was established using a randomization code in Matlab software which was shown in Appendix G. The code was run for each participant separately and numbers 1 to 15 were randomized. The numbers 1 to 5 were chosen as lean angle 5.5, the numbers 6 to 10 were chosen as 6.5 and the numbers 11 to 15 were chosen as 7.5 degrees. Before each trial, the lean angle was adjusted according to the next trial angle. The participants were instructed to recover from falling without taking any step until they feel like they don’t have any other option. After collecting fifteen trials, the harness and the markers were removed from the subjects. The anatomical measurements were performed at the end. Subject data sheet was used to record all necessary information (Appendix D).

After collecting all trials from all subjects, data were post processed using EVaRT software. The marker gaps were filled with the appropriate filling technique and then the data were smoothed with 6 cut off frequency, using butterworth filter. After filling all the gaps, the trials were exported as c3d files to be used in Visual3D software (c-motion Inc.). Subject specific skeleton models were created using subjects' anatomical measures as described in the documentation part of the Visual 3D software. Then, the necessary data were exported as text files to calculate the kinetic and kinematic variables. Matlab software was used to do all mathematical calculations. The codes that were used for data analysis were shown in Appendix F. The comparisons were made between angular momentum of the arms for 5.5 and 6.5 degrees. The angular momentum was also normalized for each participant to eliminate the effect of body mass. Body mass normalized angular momentum of the arms was analyzed.

3.4. Statistical Analyses

Statistical analyses were performed using IBM SPSS 18. Due to the limited number of participant, I used Wilcoxon Summed Ranked test to see the differences between 5.5 and 6.5 degrees in Angular Momentum variables, Ground Reaction Forces, Angular Velocities, Steady State Times, Shoulder Torques and number of recoveries. The p value was selected as .05.

I also used Linear Regression analysis to see the relationship between the physical characteristics; body mass, stature, BMI and moment of inertia, and dependent variables such as Angular Momentums, Angular velocities, Shoulder Torques and Ground Reaction Forces.

CHAPTER IV

RESULTS

This chapter, which was divided into three main sections, presents the results of the study. The first section, describes the demographic information. The second section provides data related to demographic information and number of recoveries for each participant. The last section gives the results of performance variables.

4.1. Participants

Descriptive statistics for the participants were shown in table 1

Table 2. Physical characteristics of participants

	N #	Age (yrs)	Height (m)	Weight (kg)	BMI (kg/m^2)
Male	6	23.7 ± 1.8	1.77 ± .08	87.0 ± 19.7	27.7 ± 4.8
Female	4	24.4 ± 2.4	1.68 ± .07	57.6 ± 3.8	20.4 ± 1.3
Total	12	24.0 ± 2.0	1.73 ± .08	74.8 ± 21.1	24.7 ± 5.2

4.2. Characteristics and number of recoveries of each participant

Table 3. Physical characteristics and recovery numbers of the participants

Subject	Gender	Age (yrs)	Height (m)	Weight (kg)	BMI (kg/m^2)	Recovery #	
						5.5 degrees	**6.5 degrees**
1	Male	23	1.73	76.4	25.5	4	2
2	Male	22	1.66	67.3	24.4	5	5
3	Male	25	1.86	91.4	26.4	4	1
4	Male	21	1.85	117	34.2	5	3
5	Female	27	1.79	62	19.4	4	4
6	Female	26	1.62	52	19.8	5	5
7	Male	24	1.72	71	24.0	5	2
8	Female	23	1.61	56	21.6	5	4
9	Male	25	1.77	76	24.3	5	2
10	Female	25	1.65	60	22.0	5	2
Total		**24.0 ± 2.0**	**1.73 ± .08**	**74.8 ± 21.1**	**24.7 ± 5.2**	**47**	**30**

4.3. Performance Variables

4.3.1. Number of recoveries for 5.5 and 6.5 degrees

Wilcoxon Signed Ranked Test results showed that the differences between recovery number for 5.5 and 6.5 degrees were significant (z = -2.414, p = .016) (figure 16).

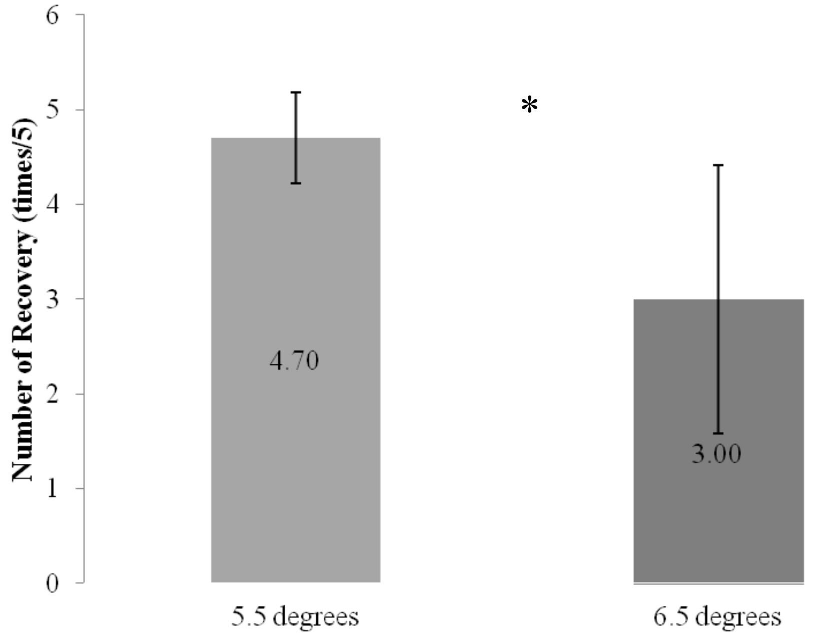


Figure 16. Recovery Number differences between 5.5 and 6.5 degrees

4.3.2. Negative Peak Angular Momentum Results (wrt Shoulder Joint)

Wilcoxon Signed Ranked Test results showed that the differences between peak negative angular momentums (H Arms) for 5.5 and 6.5 degrees were significant for both right (z = -2.803, p = .005) and the left (z = -2.805, p = .005) arms (figure 17).

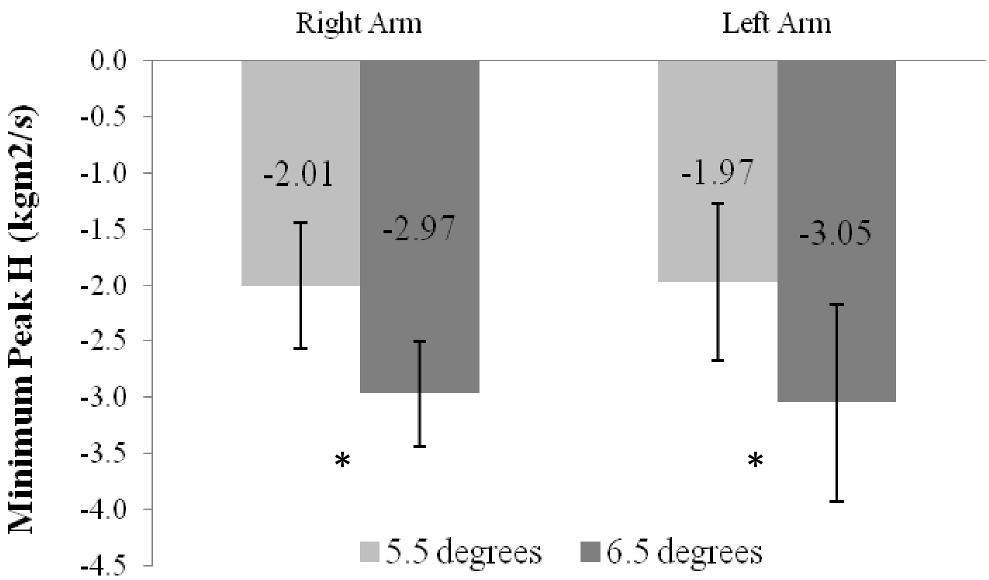


Figure 17. Negative peak angular momentum differences

between 5.5 and 6.5 degrees for right and the left arms

4.3.3. Negative Peak Angular Momentum Body Mass Normalized Results (wrt Shoulder Joint)

Wilcoxon Signed Ranked Test results showed that the differences between peak negative body mass normalized angular momentums (H Arms) for 5.5 and 6.5 degrees were significant for both right (z = -2.810, p = .005) and the left (z = -2.805, p = .005) arms (figure 18).

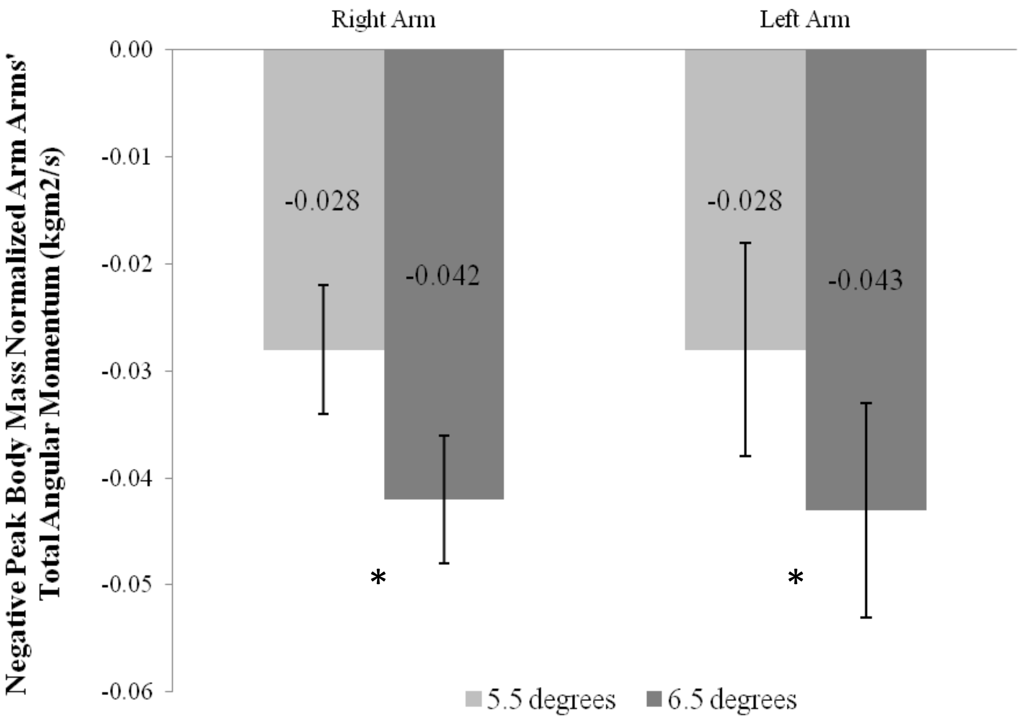


Figure 18. Negative peak body mass normalized angular momentum differences between 5.5 and 6.5 degrees for right and the left arms

4.3.4. Positive Peak Angular Momentum Results (wrt Shoulder Joint)

Wilcoxon Signed Ranked Test results showed that the differences between peak positive angular momentums (H Arms) for 5.5 and 6.5 degrees were not significant for both right (z = -.968, p = .333) and the left (z = -.153, p = .878) arms (figure 19).

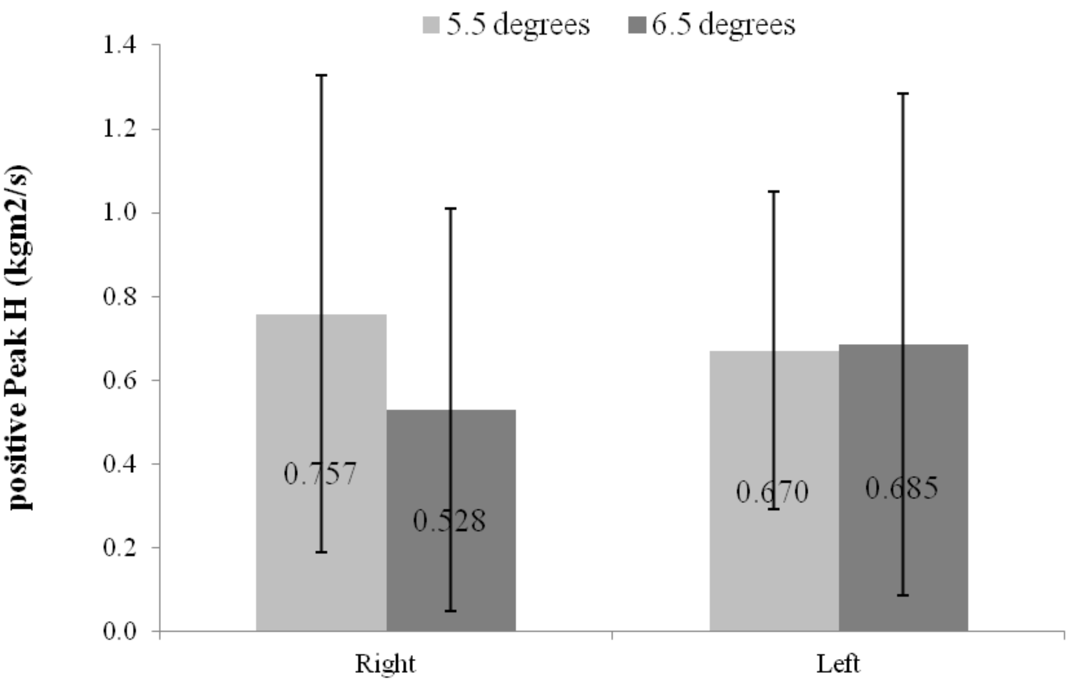


Figure 19. Positive peak angular momentum differences between 5.5 and 6.5 degrees for right and the left arms

4.3.5. Positive Peak Body Mass Normalized Arms' Total Angular Momentum Results (wrt Shoulder Joint)

Wilcoxon Signed Ranked Test results showed that the differences between peak positive body mass normalized angular momentums (H Arms) for 5.5 and 6.5 degrees were not significant for both right (z = -1.020, p = .308) and the left (z = -.051, p = .959) arms (figure 20).

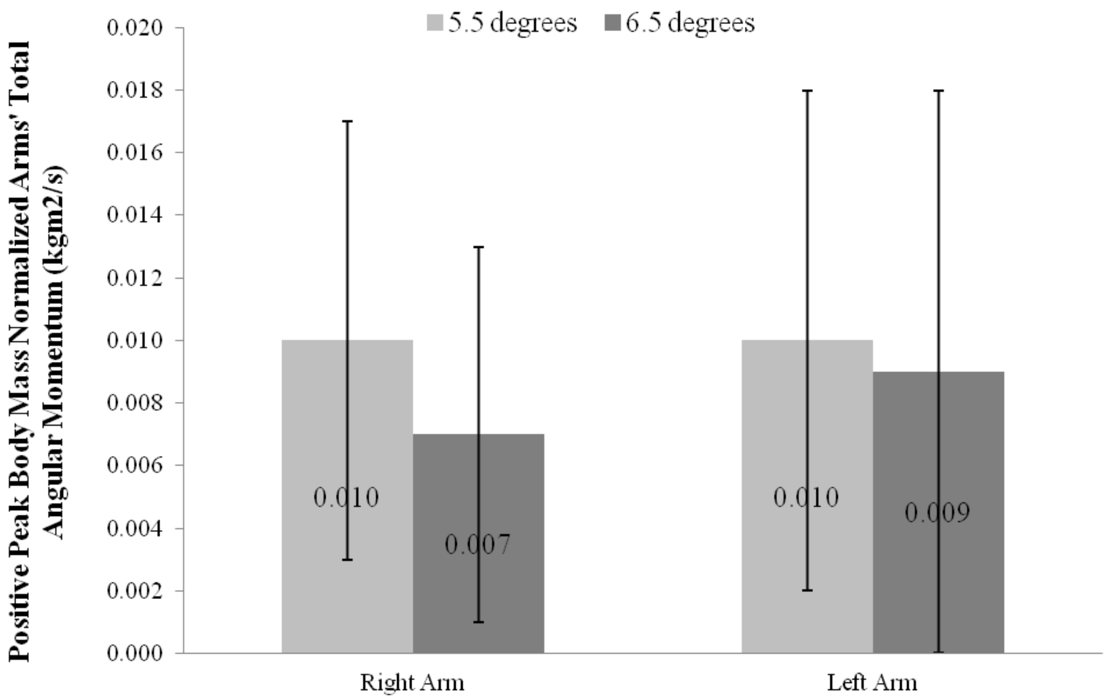


Figure 20. Positive Peak Body Mass Normalized Angular Momentum Differences

4.3.6. Right Forearm Angular Velocity differences between 5.5 and 6.5 degrees

Wilcoxon Signed Ranked Test results showed that the differences between Angular Velocities for 5.5 and 6.5 degrees were not significant (z = -2.803, p = .005) (figure 21).

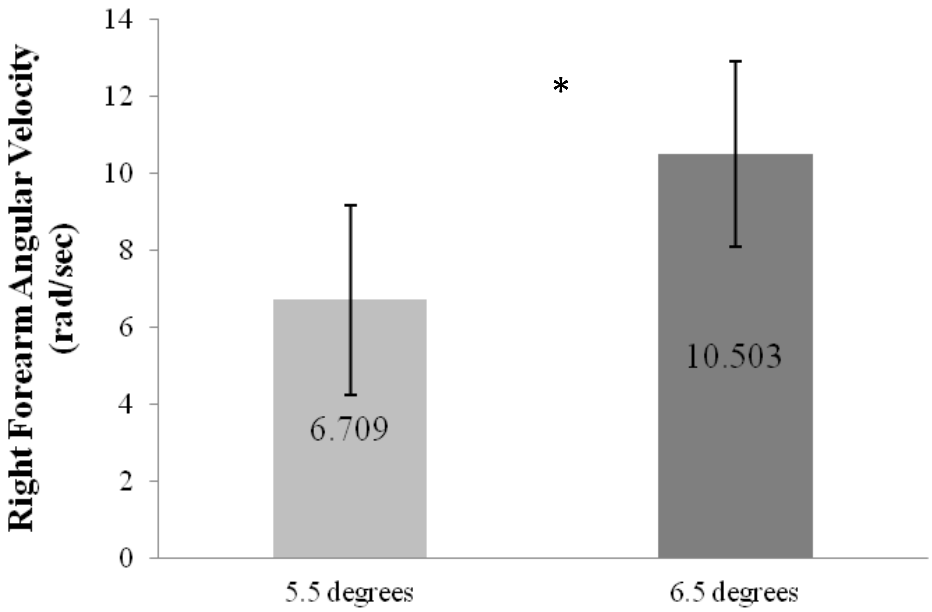


Figure 21. Right Forearm Angular Velocity differences between 5.5 and 6.5 degrees

4.3.7. Negative Peak Angular Momentum Results (wrt MTP Joint)

Wilcoxon Signed Ranked Test results showed that the differences between negative peak angular momentum of sum of both arms for 5.5 and 6.5 degrees were significant (z = -2.803, p = .005) (figure 22).

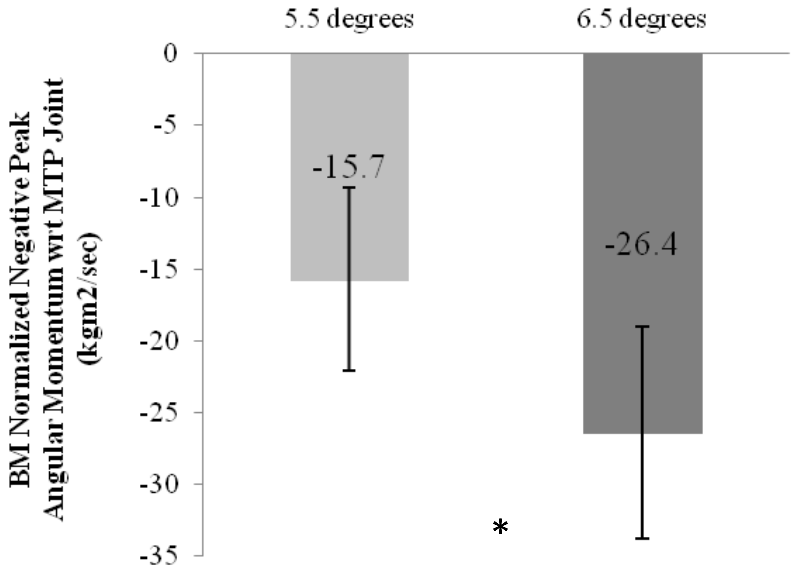


Figure 22. Negative peak angular momentum for the arms wrt MTP Joint

4.3.8. Positive Peak Angular Momentum Results (wrt MTP Joint)

Wilcoxon Signed Ranked Test results showed that the differences between negative peak angular momentum of sum of both arms for 5.5 and 6.5 degrees were significant (z = -1.988, p = .047) (figure 23).

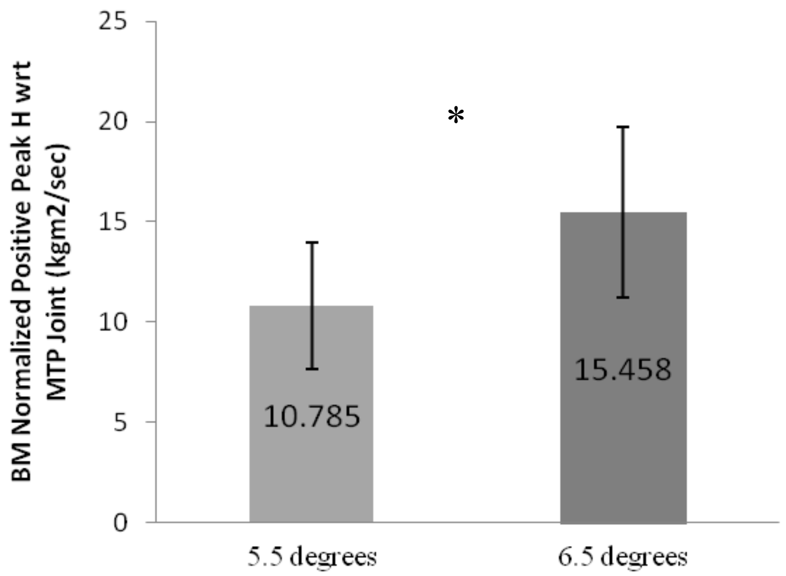


Figure 23. Positive peak angular momentum for the arms wrt MTP Joint

4.3.9. Ground Reaction Force differences between 5.5 and 6.5 degrees

Wilcoxon Signed Ranked Test results showed that the differences between GRF for 5.5 and 6.5 degrees were not significant (z = -.764, p = .445) (figure 24).

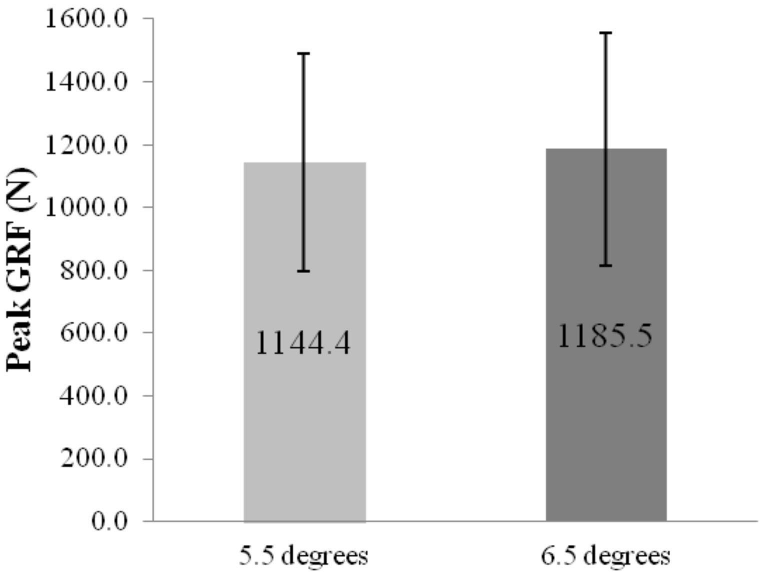


Figure 24. GRF differences between 5.5 and 6.5 degrees

4.3.10. *Steady State Time differences between 5.5 and 6.5 degrees*

Wilcoxon Signed Ranked Test results showed that the differences between Steady State Time for 5.5 and 6.5 degrees were not significant (z = -1.886, p = .059) (figure 25).

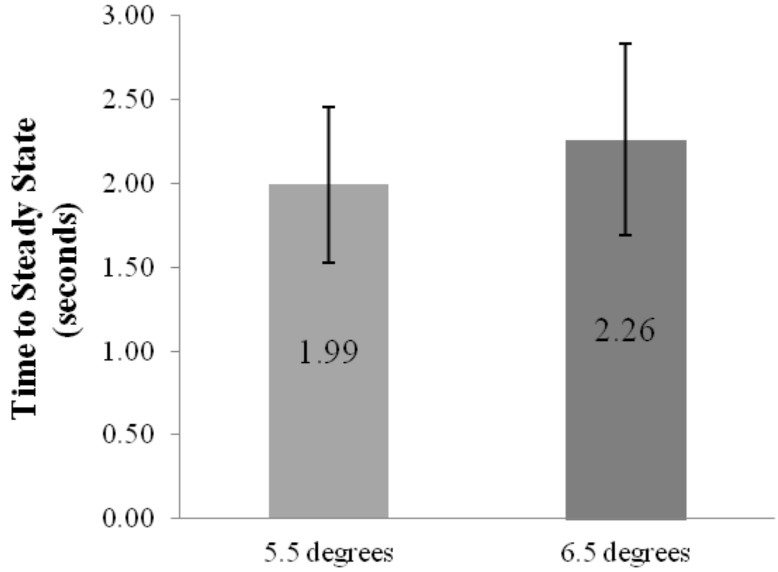


Figure 25. Steady State Time differences between 5.5 and 6.5 degrees

4.3.11. Maximum Positive Right Shoulder Torque for 5.5 and 6.5 degrees

Wilcoxon Signed Ranked Test results showed that the differences between maximum positive right shoulder torque for 5.5 and 6.5 degrees were significant (z = -2.803, p = .005) (figure 26).

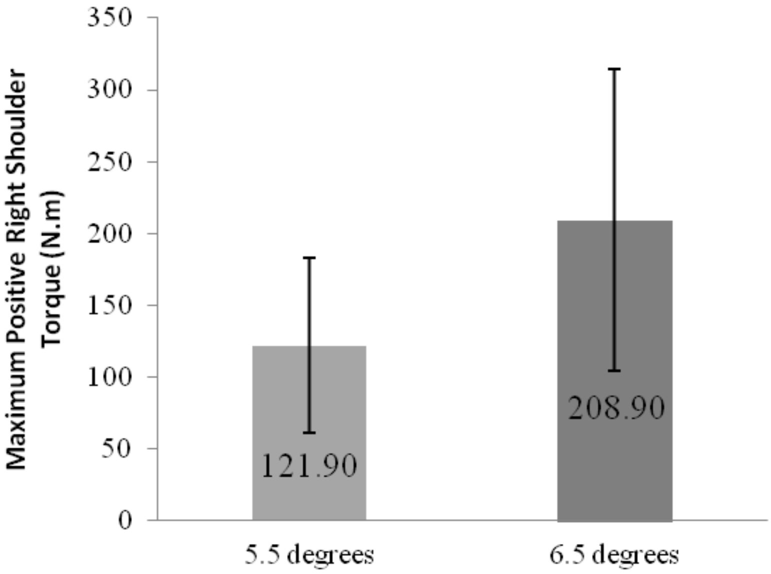


Figure 26. Maximum Positive Right Shoulder Torque for 5.5 and 6.5 degrees

4.3.12. Minimum Negative Right Shoulder Torque for 5.5 and 6.5 degrees

Wilcoxon Signed Ranked Test results showed that the differences between Negative Peak Shoulder Torque values for 5.5 and 6.5 degrees were not significant (z = -1.886, p = .059) (figure 27).

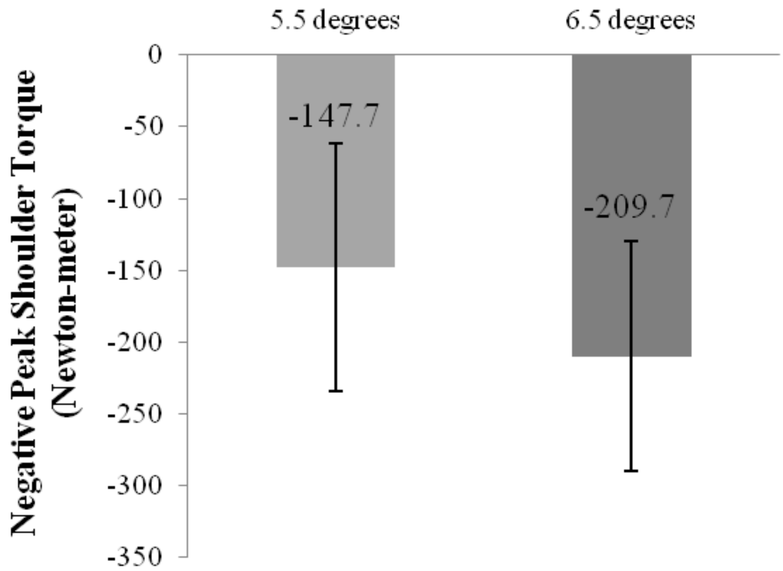


Figure 27. Minimum Negative Right Shoulder Torque for 5.5 and 6.5 degrees

4.3.13. Peak Forearm Angular Velocity (rad/sec) vs Body Mass (kg) Correlation

It was found that as the body size increases the peak forearm angular velocity decreases. The correlation in 6.5 degrees found to be significant (r squared= 0.83) (figure 28).

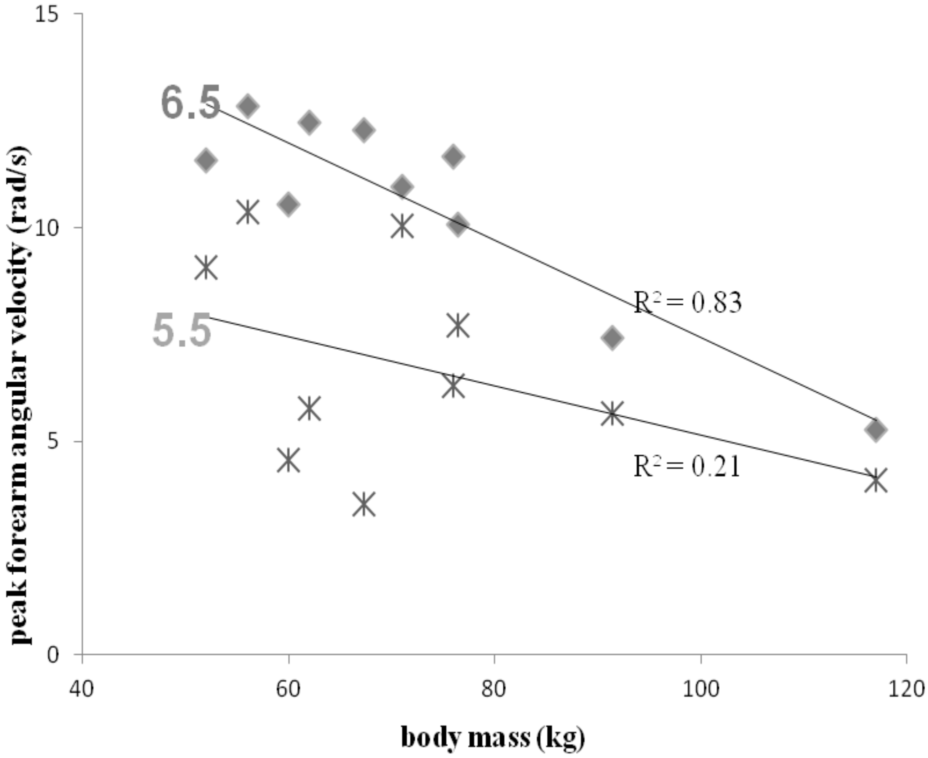


Figure 28. Peak Forearm Angular Velocity vs Body Mass Correlation

4.3.14. Peak Forearm Angular Velocity (rad/sec) vs Height (m)

.

The relationship between forearm angular velocity and height found to be correlated in 6.5 degrees which means that as the height increases the angular velocity of the arms decreases (figure 29).

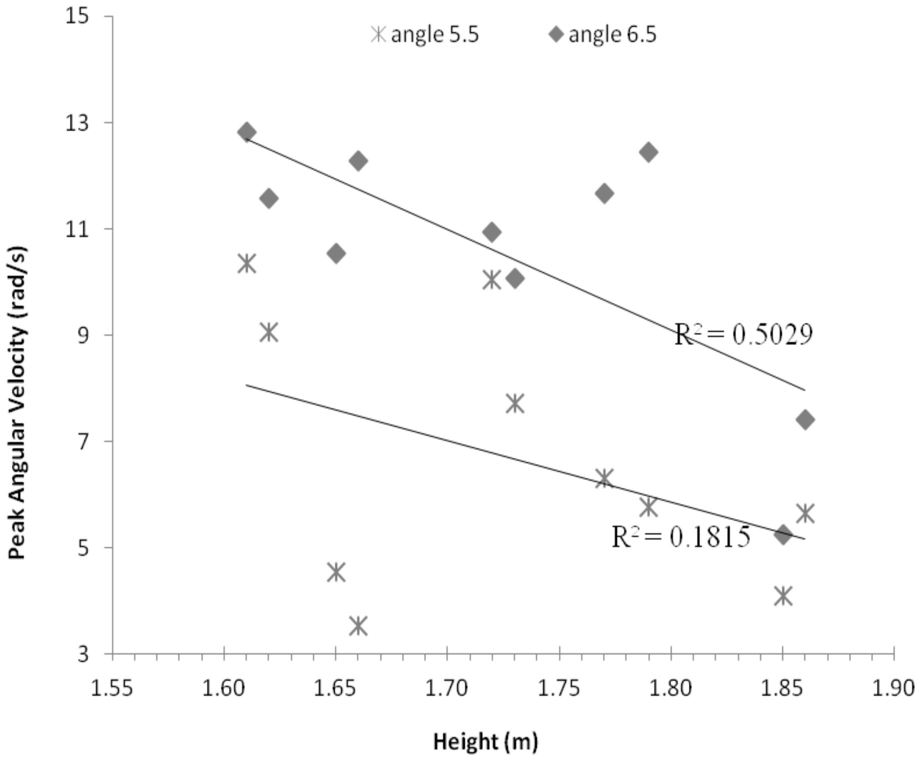


Figure 29. Peak Forearm Angular Velocity vs Height

4.3.15. Peak Forearm Angular Velocity vs BMI (kg/m2)

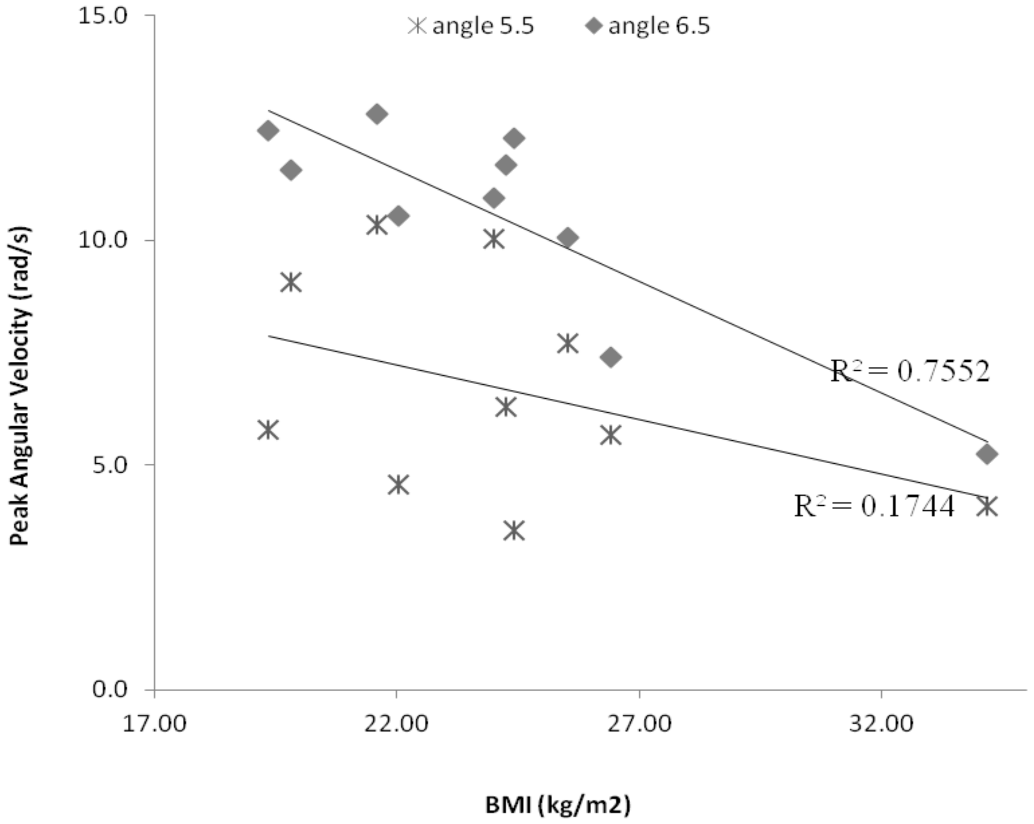


Figure 30. Peak Forearm Angular Velocity vs BMI (kg/m2)

4.3.16. Peak Forearm Angular Velocity vs Forearm Moments of Inertia

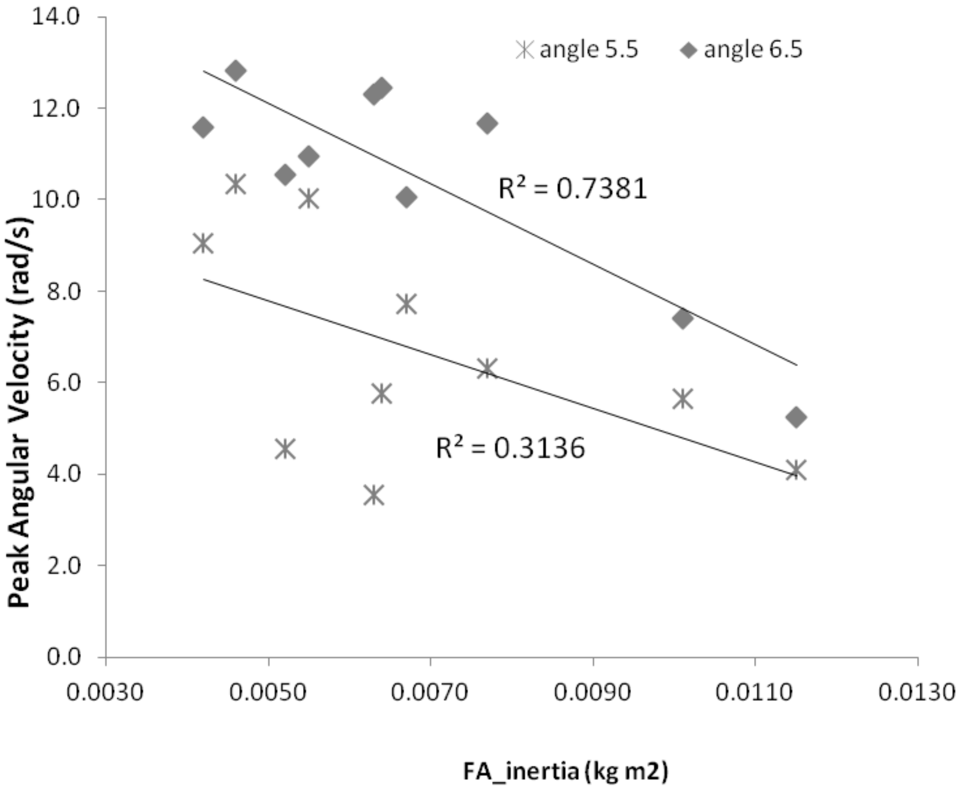


Figure 31. Peak Forearm Angular Velocity vs Forearm Moments of Inertia

CHAPTER V

DISCUSSION

The goal of this study was to understand whether the severity of forward perturbation has an effect on arm recovery strategies. 5.5, 6.5 and 7.5 degrees were used to simulate forward fall. The participants tried recovering by using their arms before taking any step. All participants used forward arm rotations, but different strategies were seen, including slow, fast, small and big circled arm rotations. None of the participants were able to recover from falling from 7.5 degrees.

Differences between numbers of recoveries for different lean angles were analyzed. Since none of the subjects were able to recover from 7.5 degrees, only 5.5 and 6.5 degrees were analyzed. Number of recoveries for 5.5 degrees was 47 out of 50. For 6.5 it was 30 out of 50. It is not surprising that participants recovered more from 5.5 degrees but this shows that the lean angle was set precise and recovery from 6.5 degree really depends on the recovery strategy. Since I do not have EMG data, I don't have a chance to understand the amount of contributions of muscle contractions especially on the ankle, knee and the hip, but I will be discussing different kinematical variables that were produced by the arms later on.

The angular momentum of the arms with respect to shoulder joint was calculated and comparisons were made for 5.5 and 6.5 degrees. Both positive and negative angular momentums were extracted. The first reaction of the arms after the initiation of perturbation was backward arm extension. This movement creates a positive angular momentum due to the laboratory coordinate system. The results showed that there were no significant differences between the 5.5 and 6.5 degrees of perturbation for positive peak angular momentum of the arms about the shoulder joint. This result was consistent for both right and the left arm. This is similar to what other researchers had also reported (McIlroy & Maki, 1995). The arms assist in shifting the whole body center mass in the posterior direction after a forward perturbation, which makes the body to move anteriorly. This opposite movement to the side of the falling is consistent with counterweight strategy. This was also observed in other studies (Marigold, Bethune, & Patla, 2003; Misiaszek, 2003). One specific possibility is that the vestibular apparatus and nuclei are involved in initiating the arm activity. Certainly, there is evidence of the importance of vestibulospinal reflex contributions to the control of the early 'automatic' postural responses of the lower limbs and axial muscles (Allum, Keshner, Honegger, & Pfaltz, 1988; Horak, Nashner, & Diener, 1990; Keshner, Woollacott, & Debu, 1988) which may even extend to amplitude modulation of the evoked reactions.

The second and the important action is the forward rotation of the arms, which creates a negative angular momentum again using the right thumb rule and the laboratory coordinate system. This movement has not been investigated before. Almost all of the research that has been conducted have focused on the early reactions of the support limb. The results of the statistical analysis showed that the differences between the 5.5 and 6.5 degree in negative angular momentum with respect to shoulder joint are significant. There is a 50% difference, strongly supporting the conclusion that subjects use their arms differently under the two conditions. The 'big picture' we were trying to see was how much processing goes on when a movement is planned. Some people believe that dynamics (such as equations of motion) are too complicated to be represented within the nervous system. However, the results of this study did not confirm this. The participants did modulate their arm swinging (ie, they did not swing as fast as they can until recovery). Both raw and normalized data showed the same results. That is, the negative angular momentum of the arms in 6.5 degrees is statistically higher than 5.5 degrees. It can be said that the arms were not rotated as a generic movement only, but they were modulated depending on the severity of perturbation. This finding was also confirmed by other studies (Daniel S. Marigold, 2003; Marigold et al., 2003; McIlroy & Maki, 1995).

The angular momentum of the arms with respect to shoulder joint gives important information about the arm modulation in different conditions but it is also important to see what arms are doing with respect to the ground. The Angular Momentum of the arms with respect to MTP joint was calculated. The participants did not move their feet and tried to recover without taking any step. Therefore, the MTP joint was the fixed point. Nonparametric Rank Summed Wilcoxon Test was used with lean angle as the only factor (with two levels: 5.5 and 6.5 degrees) and raw H_arms (Angular Momentum of the Arms) as dependent variable and found no significant effect on the first, positive peak ($p=0.347$) but it had a big effect on the second, negative peak ($p=0.008$). The first positive angular momentum can be defined as the first rotation of the arms to the back of the body, which has an impact on keeping the COM within the base of support. On the other hand, the second phase, which is the positive angular momentum, is the rotation of the arms in forward direction. This movement helps body to apply an extra force with the foot, which can also be seen in the GRF data. When the arms are rotated in front of the body counter clockwise when viewed from left, the GRF increases creating extra force to keep the body within the base of support. The difference between the 5.5 and 6.5 degrees can also be seen in shoulder torque results. The arms did generate more torque in 6.5 degrees, which explains the modulation of the arms with respect to the severity of perturbation.

After finding significant differences between 5.5 and 6.5 degrees in terms of the angular momentum, the next question we tried to answer was where these differences come from. The angular momentum can be calculated using segment mass, the distance between the segment CoM and the point of interest, the moments of inertia and angular velocities of the segments. The first question was whether the participants rotated their arms with different angular velocities. The statistical analysis showed that the participants rotated their arms significantly faster in 6.5 degrees. This result also explains the fact that the arms were not rotated as fast as possible but the participants did adjustments to recover from falling depending on the severity of perturbation.

The steady state times for 5.5 and 6.5 degrees were similar. The participants spent more time to recover from 6.5 but the difference was not high enough to be statistically significant.

When we analyzed the peak forearm angular velocity and checked whether the participants modulated their arms depending on their body mass, we saw a high dependency on their body mass. Larger subjects rotated their arms less rapidly. In figure 28, it can be seen that the sagittal plane component of the forearm angular velocity plotted versus body mass for each lean angle, with a stronger correlation between arm velocity and body mass for the higher lean angle, and a weak correlation for the lower lean angle. Similar correlations were found when arm velocity was regressed on stature, BMI, and arm segment moment of inertia.

The angular momentum and shoulder torque of the arms and GRF are overlapping and highly correlated. The differences between GRF's in 5.5 and 6.5 degrees are not high enough to show statistical significance but it can be seen that the GRF in 6.5 degrees is higher than in 5.5 degrees.

Most subjects successfully used rapid arm rotations to affect balance recovery. Graded response suggests that the degree of perturbation is taken into account when planning recovery, i.e, arms were not rotated as fast as possible until balance was recovered. Subjects with greater arm inertias compensated by rotating arms more slowly, possibly due to a less need for rapid rotation, or because more massive limbs are harder to accelerate. In conclusion, we found that most of our subjects were able to use arm rotations to recover from a potential fall.

The way that they did this depended upon the risk of falling, as indicated by the initial lean angle before release. Even though the difference between the two lean angles was only one degree, subjects generated significantly more angular momentum when released from a 6.5 degree lean than when they were released from a 5.5 degree lean. This finding implies that the arms are not rotated as fast as possible until the recovery is accomplished and that arm rotations are modulated according to the extent of the balance perturbation.

Subjects with larger arms rotated their arms more slowly. This could be attributed to the subject's modulating arm rotations according to the arm moment of inertia. That is, more massive limbs may not need to be rotated as fast to perform the same recovery. An alternative explanation is that the size of the limbs is a limiting factor in how fast the arms could be rotated.

This research suggests that arm reaction play an important role in recovering from postural perturbations by keeping the whole body center of mass within the base of support and creating more time to grasp somewhere around. Therefore, arm reaction strategies can be practiced by people to create more conscious movements to prevent falling.

The arms movements play a very crucial role in most of the sports in maintaining balance. It's very important not only to keep the body stable and perform a perfect technique but also to get ready in a very short time for the next action. For some sports that require high level of agility it is very important to be in balance. Therefore, athletes should be able to use their limbs in a controlled manner. Some of the early reactions may be generic, but they can also be practiced and improved for better performance. The angular momentum is a very important variable in maintaining the balance. The results of this research also supports the idea of central nervous system uses a very complex computation model to use different body parts to keep the body balanced.

In the future, we would like to use similar methodology to examine how the arms are used to recover from falls, initiated using different balance perturbations, such as a displacement applied to the feet, and how the arms are used to recover from loss of balance during locomotion. Regarding the next steps, we could explore this further by forcing the nervous system to perform even more complex calculations based on new information. How would people try to recover using their arms, if weights were strapped to the wrists? What sorts of compensatory movements are performed in sports to avoid falling?

Sports, such as gymnastics, figure skating, tennis and volleyball, etc. have many actions that require high level of computations to be done with in the central nervous system. A tennis player hits a backhand shot with the right arm and at the same time extends the left arm in the opposite direction to keep the angular momentum of the body close to zero so that it does not rotate to the direction of the shot. The volleyball players use their upper body and the arms while executing a spike. They also flex their lower body in the opposite direction not to roll over and fall down. The results of this research also support the idea that these counter movements are somehow controlled by central nervous system.

CHAPTER VI

SUMMARY, CONCLUSION & RECOMMENDATIONS

6.1. Summary

The objective of the present study was to understand whether the arm angular momentum is modulated depending on the severity of perturbation.

Tether release method was used to simulate a forward perturbation by putting the participants at initial lean angles of 5.5, 6.5 and 7.5 degrees. The participants were stood on a force plate and suddenly released. They were asked to recover from forward falling by using their body parts before taking any step. Reflective markers were placed on their body to collect kinematic data to compute the whole body angular momentum. The data were post processed and subject specific models were created using Visual 3d software. The segment center of mass positions, angular velocities, center of pressure, segment angles, segment velocities, and ground reaction forces were extracted. A self written Matlab code was used to analyze data.

The results showed that people are modulating their arms in different conditions. It was found that the arm angular velocity, peak negative angular momentum which is directed to the ground, and shoulder torques are higher in 6.5 degrees compared to 5.5. None of the subjects was able to recover from 7.5 degrees, which was also supported by the literature. The maximum lean angle that can be recovered was found to be 7.2 degree by young people. It was also found that correlation between body mass and angular velocities, which shows that the heavy participants do not need to rotate their arms faster since they can generate the same angular momentum with slower rotations.

6.2. *Conclusion*

In conclusion, it can be speculated that the participants of this study did not use generic or reflexive movements when they face with an unexpected balance perturbation. The central nervous system somehow does some calculations to determine the amount of action that is necessary to recover from falling. The arms are not rotated as fast as possible until the person recovers but the rotation of the arms is modulated depending on the severity of perturbation. However, more research is needed to understand mechanisms used to recover from falling.

6.3. *Recommendations*

Up to date, literature focused on different types of perturbations and tried to understand what people are doing with their muscles, what sort of stepping actions are taken. This is the first study to understand the effects of arms in recovering from postural perturbations. Moreover, this is the only study that focused on arms rotations. To extend this study, our plan is to create an inverted pendulum simulation model to understand the effects of different variables in creating different amounts of angular momentum not only for balance recovery but also for different sporting actions.

The second study we are planning is to strap extra weights on the arms and see how it affects the recovery strategies.

Considering the sports, volleyball has been the field we want to extend our research. During a spike action the volleyball player hits the ball by flexing the arms and upper body forcefully. If the player does not do an opposite action that is necessary to equalize the angular momentum the player cannot prevent rolling over and land in a risky way probably causing an injury. To prevent this, lower body is flexed and the angular momentum kept zero, so that the player does not lose balance in the air and land on the floor in a secured manner. The question is what happens if we strap some extra weight on the ankles. Using the same principle does the player generate extra force with the upper body and hit the ball stronger?

More room is available for other sports such as gymnastics, track and field throwing events. What sorts of compensatory movements are performed in sports to avoid falling and improve the performance?

What if we strap some extra weight on the non dominant arm of a tennis player when he is executing a backhand shot. Would the player hit the ball faster to equalize the angular momentum with the racket hand?

REFERENCES

Allum, J. H., Keshner, E. A., Honegger, F., & Pfaltz, C. R. (1988). Organization of leg-trunk-head equilibrium movements in normals and patients with peripheral vestibular deficits. *Prog Brain Res, 76*, 277-290.

Ananth, L. (2014). Cerebellum Balance & Motor Control in Brain. 2014, from https://suite101.com/a/cerebellum-balance-motor-control-in-brain-a167209

Arus, E. (2012). *Biomechanics of human motion applications in the martial arts*: Taylor and Francis Group.

Blake, A. J., Morgan, K., Bendall, M. J., Dallosso, H., Ebrahim, S. B. J., Arie, T. H. D., . . . Bassey, E. J. (1988). Falls by Elderly People at Home - Prevalence and Associated Factors. *Age and Ageing, 17*(6), 365-372. doi: 10.1093/ageing/17.6.365

Bothner, K. E., & Jensen, J. L. (2001). How do non-muscular torques contribute to the kinetics of postural recovery following a support surface translation? *J Biomech, 34*(2), 245-250. doi: S0021-9290(00)00161-5 [pii]

Carpenter, M. G., Allum, J. H., & Honegger, F. (1999). Directional sensitivity of stretch reflexes and balance corrections for normal subjects in the roll and pitch planes. *Exp Brain Res, 129*(1), 93-113. doi: 91290093.221 [pii]

Cham, R., & Redfern, M. S. (2001). Lower extremity corrective reactions to slip events. *Journal of Biomechanics, 34*(11), 1439-1445. doi: http://dx.doi.org/10.1016/S0021-9290(01)00116-6

CIHI. (2013). National Trauma Registry Report 2013: Hospitalizations for Major Injury in Canada, 2010-2011.

Cumming, R. G., & Klineberg, R. J. (1994). Fall Frequency and Characteristics and the Risk of Hip-Fractures. *Journal of the American Geriatrics Society, 42*(7), 774-778.

Daniel S. Marigold, A. J. B., and Aftab E. Patla. (2003). Role of the Unperturbed Limb and Arms in the Reactive Recovery Response to an Unexpected Slip During Locomotion. *J Neurophysiol, 89*, 1727-1737.

Eurich, C. W., & Milton, J. G. (1996). Noise-induced transitions in human postural sway. *Phys Rev E Stat Phys Plasmas Fluids Relat Interdiscip Topics, 54*(6), 6681-6684.

Forner Cordero, A., Koopman, H. F., & van der Helm, F. C. (2003). Multiple-step strategies to recover from stumbling perturbations. *Gait Posture, 18*(1), 47-59. doi: S0966636202001601 [pii]

Gabell, A., Simons, M. A., & Nayak, U. S. L. (1985). Falls in the Heathy Elderly - Predisposing Causes. *Ergonomics, 28*(7), 965-975. doi: 10.1080/00140138508963219

Hayes, W. C., Myers, E. R., Robinovitch, S. N., VandenKroonenberg, A., Courtney, A. C., & McMahon, T. A. (1996). Etiology and prevention of age-related hip fractures. *Bone, 18*(1), S77-S86. doi: 10.1016/8756-3282(95)00383-5

Henry, S. M., Fung, J., & Horak, F. B. (1998). EMG responses to maintain stance during multidirectional surface translations. *J Neurophysiol, 80*(4), 1939-1950.

Hilliard, M. J., Martinez, K. M., Janssen, I., Edwards, B., Mille, M.-L., Zhang, Y., & Rogers, M. W. (2008). Lateral Balance Factors Predict Future Falls in Community-Living Older Adults. *Archives of Physical Medicine and Rehabilitation, 89*(9), 1708-1713. doi: http://dx.doi.org/10.1016/j.apmr.2008.01.023

Horak, F. B., & Nashner, L. M. (1986). Central programming of postural movements: adaptation to altered support-surface configurations. *J Neurophysiol, 55*(6), 1369-1381.

Horak, F. B., Nashner, L. M., & Diener, H. C. (1990). Postural strategies associated with somatosensory and vestibular loss. *Exp Brain Res, 82*(1), 167-177.

Horak, F. B., Shumway-Cook, A., Crowe, T. K., & Black, F. O. (1988). Vestibular function and motor proficiency of children with impaired hearing, or with learning disability and motor impairments. *Dev Med Child Neurol, 30*(1), 64-79.

Horak, F. B., Shupert, C. L., & Mirka, A. (1989). Components of Postural Dyscontrol in the Elderly - A Review. *Neurobiology of Aging, 10*(6), 727-738. doi: 10.1016/0197-4580(89)90010-9

Hsiao-Wecksler, E. T. (2008). Biomechanical and age-related differences in balance recovery using the tether-release method. *J Electromyogr Kinesiol, 18*(2), 179-187. doi: 10.1016/j.jelekin.2007.06.007

Keshner, E. A., Woollacott, M. H., & Debu, B. (1988). Neck, trunk and limb muscle responses during postural perturbations in humans. *Exp Brain Res, 71*(3), 455-466.

Luchies, C. W., Alexander, N. B., Schultz, A. B., & Ashton-Miller, J. (1994). Stepping responses of young and old adults to postural disturbances: kinematics. *J Am Geriatr Soc, 42*(5), 506-512.

M.McGinnis, P. (2013). *Biomechanics of Sport and Exercise*

Mackey, D. C. (2004). *Biomechanics of postural stability in the elderly.* (Master of Science), Simon Fraser University.

Mackey, D. C., & Robinovitch, S. N. (2006). Mechanisms underlying age-related differences in ability to recover balance with the ankle strategy. *Gait & Posture, 23*(1), 59-68. doi: DOI 10.1016/j.gaitpost.2004.11.009

Maki, B. E., Holliday, P. J., & Fernie, G. R. (1987). A posture control model and balance test for the prediction of relative postural stability. *IEEE Trans Biomed Eng, 34*(10), 797-810.

Maki, B. E., & McIlroy, W. E. (1997). The role of limb movements in maintaining upright stance: the "change-in-support" strategy. *Phys Ther, 77*(5), 488-507.

Manchester, D., Woollacott, M., Zederbauerhylton, N., & Marin, O. (1989). Visual, Vestibular and Somotosensory Contributions to Balance Control in the Older Adult. *Journals of Gerontology, 44*(4), M118-M127.

Marigold, D. S., Bethune, A. J., & Patla, A. E. (2003). Role of the unperturbed limb and arms in the reactive recovery response to an unexpected slip during locomotion. *Journal of Neurophysiology, 89*(4), 1727-1737. doi: 10.1152/jn.00683.2002

McCollum, G., Shupert, C. L., & Nashner, L. M. (1996). Organizing sensory information for postural control in altered sensory environments. *J Theor Biol, 180*(3), 257-270. doi: S0022-5193(96)90101-0 [pii], 10.1006/jtbi.1996.0101

McIlroy, W. E., & Maki, B. E. (1995). Early activation of arm muscles follows external perturbation of upright stance. *Neurosci Lett, 184*(3), 177-180. doi: 0304394094112003 [pii]

McIlroy, W. E., & Maki, B. E. (1996). Age-related changes in compensatory stepping in response to unpredictable perturbations. *J Gerontol A Biol Sci Med Sci, 51*(6), M289-296.

Mille, M. L., Johnson-Hilliard, M., Martinez, K. M., Zhang, Y. H., Edwards, B. J., & Rogers, M. W. (2013). One Step, Two Steps, Three Steps More Directional Vulnerability to Falls in Community-Dwelling Older People. *Journals of Gerontology Series a-Biological Sciences and Medical Sciences, 68*(12), 1540-1548. doi: DOI 10.1093/gerona/glt062

Misiaszek, J. (2003). Early activation of arm and leg muscles following pulls to the waist during walking. *Experimental Brain Research, 151*(3), 318-329. doi: 10.1007/s00221-003-1501-x

Morfitt, J. M. (1983). Falls in Old-People at Home - Intrinsic Versus Environmental-Factors in Causation. *Public Health, 97*(2), 115-120. doi: 10.1016/s0033-3506(83)80008-0

Moss, F., & Milton, J. G. (2003). Balancing The Unbalanced. *Nature, 425*, 911-912.

Murnaghan, C. D. (2008). *The influence of voluntary movement dynamics on postural stability borders and balance recovery strategies.* (Master of Science), Simon Fraser University.

Nashner, L. M. (1976). Adapting reflexes controlling the human posture. *Exp Brain Res, 26*(1), 59-72.

Nashner, L. M., Woollacott, M., & Tuma, G. (1979). Organization of rapid responses to postural and locomotor-like perturbations of standing man. *Exp Brain Res, 36*(3), 463-476.

Nevitt, M. C., & Cummings, S. R. (1994). Type of Fall and Risk of Hip and Wrist Fractures - The Study of Osteoporotic Fractures. *Journal of the American Geriatrics Society, 42*(8), 909-909.

Pai, Y.-C., Rogers, M. W., Patton, J., Cain, T. D., & Hanke, T. A. (1998). Static versus dynamic predictions of protective stepping following waist–pull perturbations in young and older adults. *Journal of Biomechanics, 31*(12), 1111-1118. doi: http://dx.doi.org/10.1016/S0021-9290(98)00124-9

Parker, M. J., Twemlow, T. R., & Pryor, G. A. (1996). Environmental hazards and hip fractures. *Age and Ageing, 25*(4), 322-325. doi: 10.1093/ageing/25.4.322

Pavol, M. J., Owings, T. M., Foley, K. T., & Grabiner, M. D. (1999). Gait characteristics as risk factors for falling from trips induced in older adults. *J Gerontol A Biol Sci Med Sci, 54*(11), M583-590.

Pavol, M. J., Runtz, E. F., Edwards, B. J., & Pai, Y. C. (2002). Age influences the outcome of a slipping perturbation during initial but not repeated exposures. *J Gerontol A Biol Sci Med Sci, 57*(8), M496-503.

Pijnappels, M., Bobbert, M. F., & van Dieen, J. H. (2004). Contribution of the support limb in control of angular momentum after tripping. *J Biomech, 37*(12), 1811-1818. doi: S0021929004001319 [pii], 10.1016/j.jbiomech.2004.02.038

Pijnappels, M., Kingma, I., Wezenberg, D., Reurink, G., & Dieën, J. H. (2009). Armed against falls: the contribution of arm movements to balance recovery after tripping. *Experimental Brain Research, 201*(4), 689-699. doi: 10.1007/s00221-009-2088-7

Rogers, M. W., Hedman, L. D., Johnson, M. E., Cain, T. D., & Hanke, T. A. (2001). Lateral stability during forward-induced stepping for dynamic balance recovery in young and older adults. *J Gerontol A Biol Sci Med Sci, 56*(9), M589-594.

Runge, C. F., Shupert, C. L., Horak, F. B., & Zajac, F. E. (1999). Ankle and hip postural strategies defined by joint torques. *Gait Posture, 10*(2), 161-170. doi: S0966-6362(99)00032-6 [pii]

SMARTRISK. (2004). The Economic Burden of Injury in Canada. Retrieved 13.12.2013, 2013, from http://www.phac-aspc.gc.ca/injury-bles/ebuic-febnc/

Smeesters, C., Hayes, W. C., & McMahon, T. A. (2001). Disturbance type and gait speed affect fall direction and impact location. *Journal of Biomechanics, 34*(3), 309-317. doi: http://dx.doi.org/10.1016/S0021-9290(00)00200-1

Stability and Balance. (2008). Retrieved 12.11.2013, 2013, from http://lilinfo.wordpress.com/2008/08/17/stability-and-balance/

Stelmach, G. E., Phillips, J., Difabio, R. P., & Teasdale, N. (1989). Age, Functional Postural Reflexes, and Voluntary Sway. *Journals of Gerontology, 44*(4), B100-B106.

Tokuno, C. D., Carpenter, M. G., Thorstensson, A., & Cresswell, A. G. (2006). The influence of natural body sway on neuromuscular responses to an unpredictable surface translation. *Exp Brain Res, 174*(1), 19-28. doi: 10.1007/s00221-006-0414-x

Topper, A. K., Maki, B. E., & Holliday, P. J. (1993). Are Activity-Based Assessments of Balance and Gait in The Elderly Predictive of Risk of Falling and or Type of Fall. *Journal of the American Geriatrics Society, 41*(5), 479-487.

Troy, K. L., & Grabiner, M. D. (2006). Recovery responses to surrogate slipping tasks differ from responses to actual slips. *Gait Posture, 24*(4), 441-447. doi: S0966-6362(05)00251-1 [pii], 10.1016/j.gaitpost.2005.09.009

WHO. (2012). Falls. Retrieved 12.11.2013, 2013, from http://www.who.int/mediacentre/factsheets/fs344/en/

Woollacott, M., & Shumway-Cook, A. (2002). Attention and the control of posture and gait: a review of an emerging area of research. *Gait & Posture, 16*(1), 1-14. doi: 10.1016/s0966-6362(01)00156-4

APPENDICES

A. INFORMED CONSENT FORM

PENNSTATE

Informed Consent Form for Biomedical Research

The Pennsylvania State University

ORP USE ONLY: IRB# **28978 Doc. # 1**
The Pennsylvania State University
Office for Research Protections
Approval Date: 08/12/2008 ARS
Expiration Date: 08/05/2009 ARS
Social Science Institutional Review Board

Title of Project: **Investigation of the Use of the Arms in Recovering from Postural Perturbations**

Principal Investigator: **Stephen J. Piazza**

Biomechanics Laboratory

29 Recreation Building

Penn State University

814-865-3413

piazza@psu.edu

Other Investigators: **Emre Ak**

1. **Purpose of the study:**

   You are being asked to participate in a study being conducted in The Biomechanics Laboratory, which is a laboratory specializing in studies of posture and walking. The purpose of this study is to measure the motions of your body as you try to recover from a fall.

2. **Procedures to be followed:**

   If you are healthy and between the ages of 18-25 years, you can participate in this study. You will be asked to try to recover from a fall after you lean forward and a cable supporting you is suddenly released. Your motion will be recorded using video cameras that track the motions of reflective markers placed on your body at several locations: on your feet, lower legs, thighs, pelvis, trunk, head, upper arms, and lower arms.

3. **Discomforts and risks:**

   All motions that you will be performing are within the range of normal movements. However, there is a risk that you may fall while you lean forward. This risk is reduced by selecting volunteers that are healthy and through the use of a harness you will wear that is attached to the ceiling. There is a risk that you might injure your muscles while performing some of the activities, but you will not be asked to perform any activity with which you are not comfortable. There will also be pads on the floor to reduce the risk of injury in the unlikely event that the harness system fails. Your skin may be irritated (redness, itching, etc) due to the adhesive taping of the reflective markers, but this should disappear within 24 hours.

4. a. Benefits to me:

There is no direct benefit for participating in this research.

b. Potential benefits to society:

The results of this study may contribute to understand strategies used to prevent falls.

5. Duration/time of the procedures and study:

Your visit to The Biomechanics Laboratory will last approximately two hours.

6. Alternative procedures that could be utilized:

The described procedures were designed to obtain specific types of data; no other alternatives exist. You may choose not to participate in this research.

7. Statement of confidentiality:

Any data collected in this experiment will remain confidential. The data will be located and secured within a locked room inside a file cabinet in The Biomechanics Laboratory and remain under the supervision of Dr. Stephen Piazza. The information will be kept confidential; only the investigator and his assistants will have access to the data. Any identifiers, such as your name or personal information, will be kept separate from the actual data.

The following may review and copy records related to this research: The Office of Human Research Protections in the U.S. Department of Health and Human Services, the Biomedical Institutional Review Board and the PSU Office for Research Protections.

Video recordings of this research will be made. These recordings will be transferred to a computer and stored there indefinitely, but the original video tapes will be erased immediately after the video has been transferred to the computer. Only the researchers associated with this project will have access to the computer video files.

8. Right to ask questions:

Please contact Dr. Stephen Piazza at (814) 865-3413 with questions, complaints or concerns about this research. You can also call this number if you feel this study has harmed you. Questions about your rights as a research participant may be directed to PennState University's Office for Research Protections at (814) 865-1775.

9. Voluntary participation:

Participation is voluntary. You can stop at any time. You do not have to answer any questions you do not want to answer. Refusal to take part in or withdrawing from this study will involve no penalty or loss of benefits you would receive otherwise.

10. Injury Clause:

> In the unlikely event you become injured as a result of your participation in this study, medical care is available but neither financial compensation nor free medical treatment is provided. By signing this document, you are not waiving any rights that you have against The Pennsylvania State University for injury resulting from negligence of the University or its investigators**.**

You must be 18 years of age or older to take part in this research study. If you agree to take part in this research study and the information outlined above, please sign your name and indicate the date below.

You will be given a copy of this signed and dated consent for your records.

____ ______________________

Participant Signature **Date**

____ ______________________

Person Obtaining Consent **Date**

B. RECRUITMENT SCRIPT

"Hi, my name is _______________ and I would like to know if you would be interested in participating as a subject in a research study being conducted at the Biomechanics Laboratory at Penn State University. The study would take about two hours of your time on one day and does not involve painful or invasive measurements. We are interested in measuring the motions of your body as you lean forward and then try to recover from falling. Reflective markers will be placed on your body at several locations and cameras will be used to track those markers as you perform the activities. The reason we are performing this study is to understand strategies used to recover from forward falling. To participate in this study, you must be above the age of 18, have no neurological problems or movement disorders and have no significant musculoskeletal injuries in the past year, or have difficulty in performing any of the activities we will be studying."

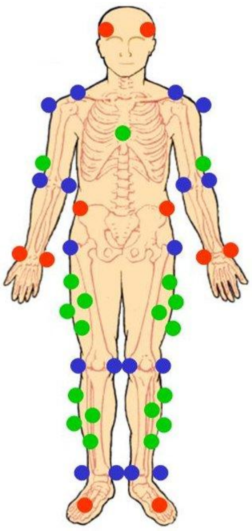
"If you have any questions at any time before the study, you can contact me at the Biomechanics Laboratory at Penn State University at ___________ or via e-mail: ____________."

C. MARKER SET GUIDELINES

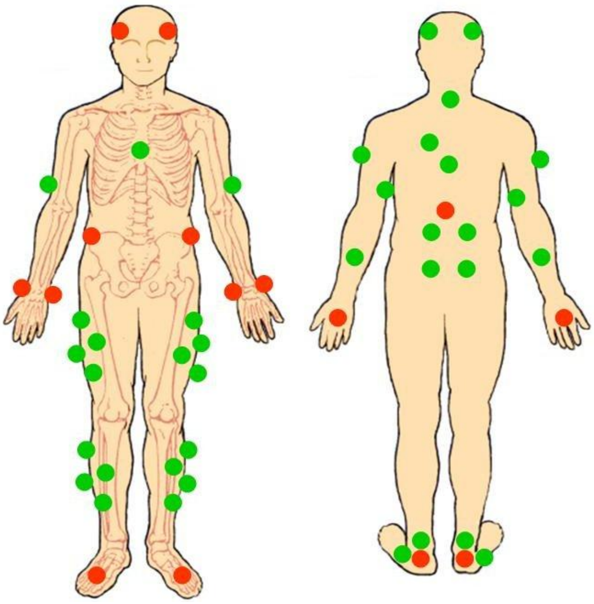
The following image displays a frontal image of a conservative full body marker set. This should be considered an ideal marker set, and there are many reasons why it might be necessary to compromise on this marker set. This topic simply provides our recommendation for ideal circumstances.

The different colored circles relate to the different roles of the markers.

Red Markers are used for both the segment definition and for tracking
Green Markers are used for tracking only
Blue markers are used only for the segment definitions (these can be removed for the movement trial)



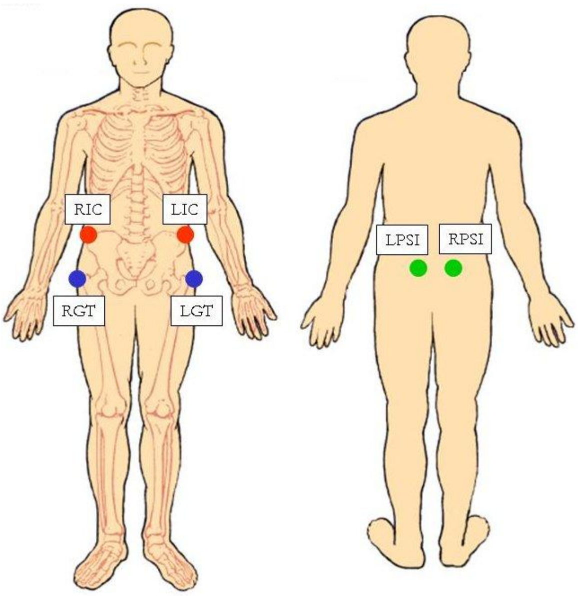
The following image shows a frontal and back view of only the markers that are used to track the segments. Note that all segments have at least 3 markers attached to them, so that it is possible to computed 6 degree of freedom segments.



The following marker names are consistent with this marker set. We recommend short marker labels because longer marker names tend to clutter the 3D viewer. There is nothing special about these marker names, however, and users are welcome to label them as it suits them.

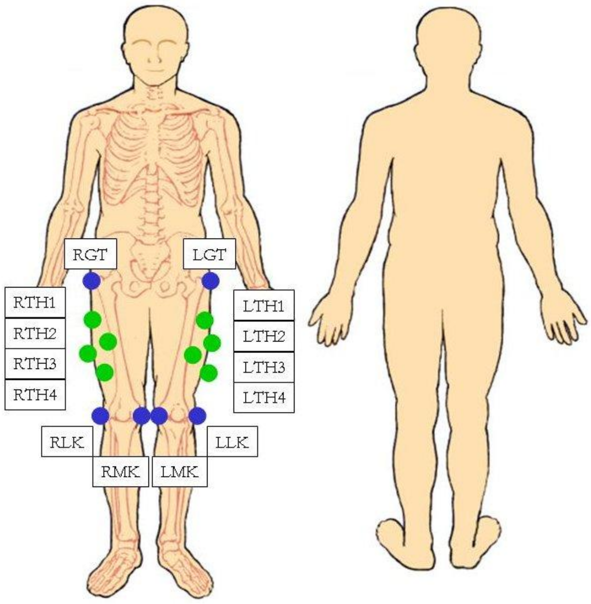
Pelvis Segment

Note that we don't recommend that ASIS markers are used. In many cases the ASIS markers are fine, but this isn't generally true.



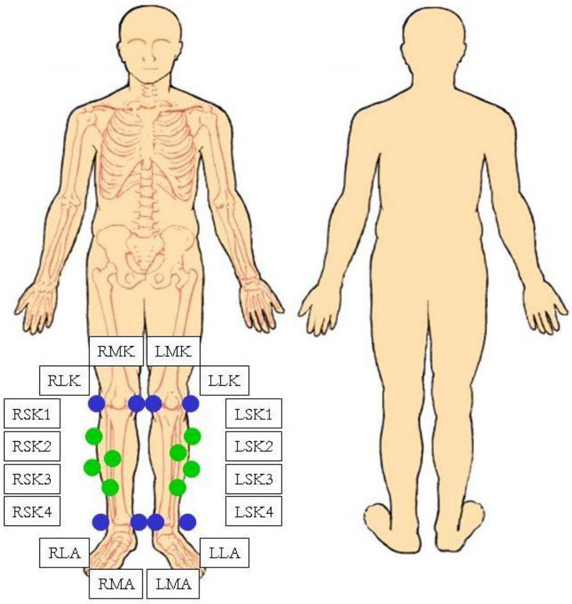
Thigh Segment

it is recommended that users avoid using greater trochanter or lateral knee markers as tracking markers for the thigh because of the inordinate amount of soft tissue artifact.



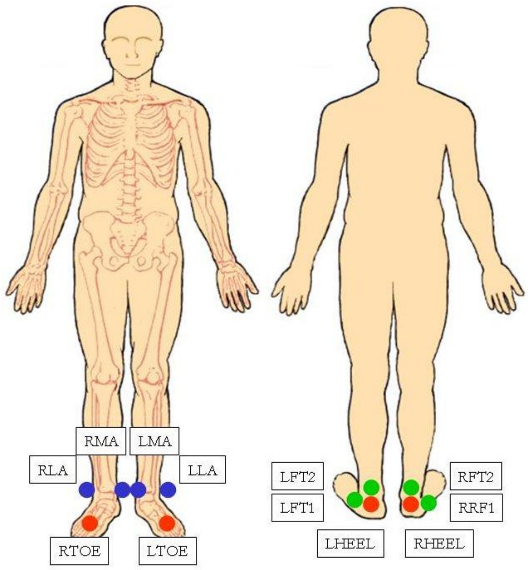
Shank Segment

Consistent with our recommendation for the thigh, we recommend that users avoid the lateral knee marker as a tracking marker.



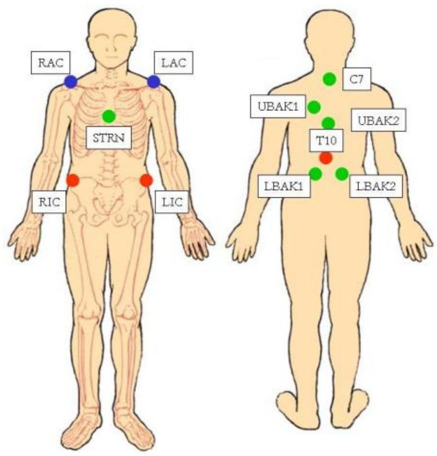
Foot Segment

This model is consistent with a one segment foot segment. Multi segment foot models (such as the Oxford Foot Model) require more markers. Note that in this example all 3 tracking markers are secured to the calcaneous. This is convenient for many studies because two markers can be easily knocked off; **which necessitates redoing the standing trial because the marker will never be re-placed exactly as it was placed**. For many studies, such as walking, a marker on the mid foot is appropriate.



Thorax/Ab Segment

This marker set is consistent with both a one segment torso (Iliac Crest to Acromium) and a two segment torso; lumbar segment from Iliac Crest to a projection of the T10 (or T12) marker onto a plane defined by the Iliac Crest and Acromium; thoracic segment from the projected T10 marker to the acromium.

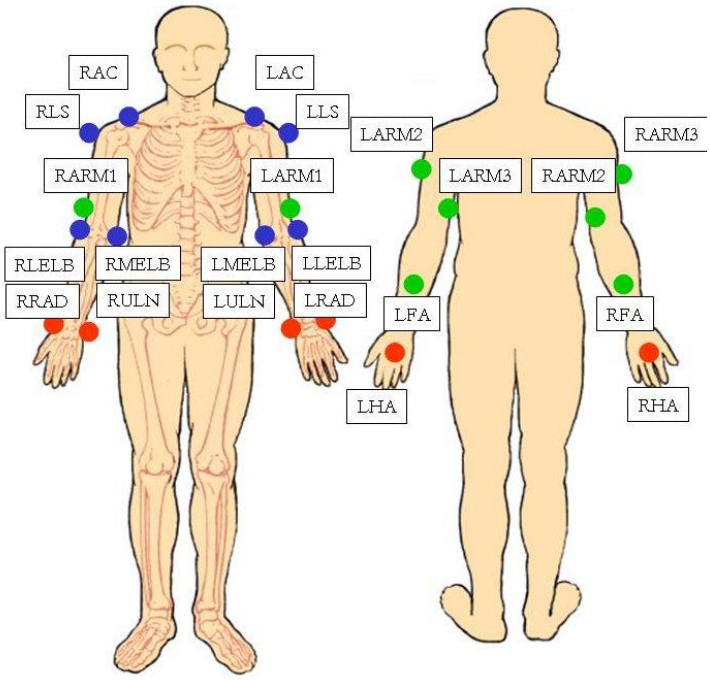


Upper Arm and Lower Arm Segment

The upper arm segment is defined from the head of the humerus (shoulder "joint") to the elbow. In this example the shoulder "joint" is offset from the acromium marker consistent with the International Shoulder Group. In most cases we recommend computing the shoulder joint using functional joints.

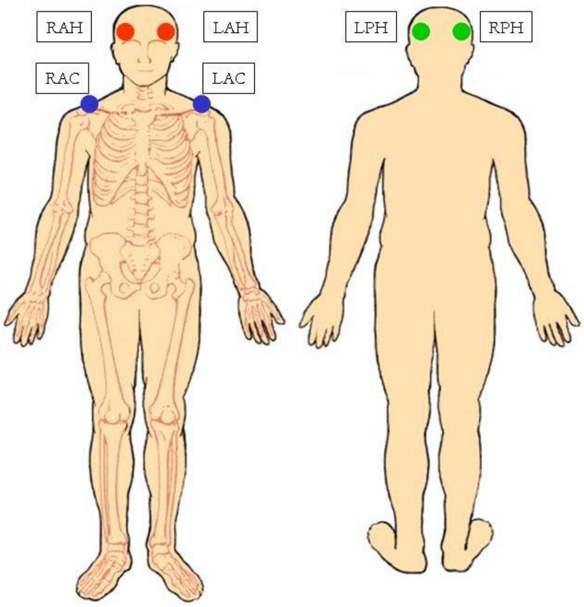
In this example, the orientation of the upper arm is determined by the placement of the medial and lateral elbow markers. It is quite difficult to place the medial marker, so we often recommend a standing posture in which the elbows are flexed at 90 degrees and the frontal plane of the upper arm is determined by the axial axis of the forearm.

The forearm segment is defined from the elbow joint (e.g. distal end of the upper arm segment) and the medial and lateral wrist markers assuming that the wrist markers are placed laterally on the wrist; if the wrist markers are placed on the posterior surface, it is necessary to create landmarks that are projected. It is often convenient to use a digitizing pointer to identify these locations. Note that supination/pronation of the forearm causes the medial and lateral elbow landmarks to NOT lie in the frontal plane of the forearm, so they should not be used as the proximal medial and lateral markers for the forearm.

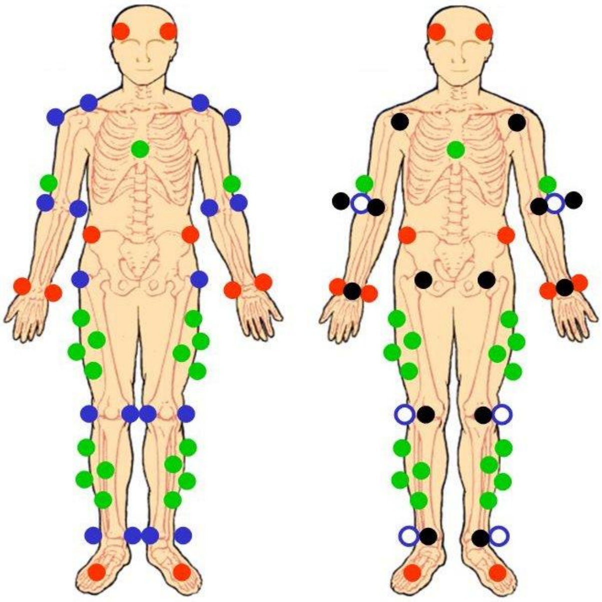


Head Segment

If the purpose of the head segment is for kinematics only, the following definition is easy to use. The proximal end of the head is at the acromium and the distal end is vertically above the acromium at the level of the ear.



The following image contains black markers (functional) and open circles (digitized) that represent locations that can be determined from functional joints and a digitizing pointer.



Marker Names associated with the images.

RAH	RPH		LAH	LPH
RLS	RAC		LAC	LLS
RARM3	RARM1	C7	LARM1	LARM3
	RARM2	UBAK1	LARM2	
RMELB	RLELB	UBKA2	LMELB	LLELB
	RFA	STRN	LFA	
RULN	RRAD	T10	LULN	LRAD
	RHA	LBAK1	LHA	
	RIC	LBAK2	LIC	
	RPSI		LPSI	
	RGT		LGT	
RTH3	RTH1		LTH1	LTH3
RTH4	RTH2		TLH2	LTH4
RMK	RLK		LMK	LLK
RSK3	RSK1		LSK1	LSK3
RSK4	RSK2		LSK2	LSK4
RMA	RLA		LMA	LLA
RFT2	RFT1		LFT1	LFT2
	RHEEL		LHEEL	
	RTOE		LTOE	

D. PARTICIPANT'S DATA SHEET

Subject ID	: S-5
Age	: 22
Weight	: 148 pound
Height	: 1.66 m
Foot Length	: 25 cm
Pelvis Depth	: 22
Trunk Depth	: 22
Knee Radius	: 5
Ankle Radius	: 3.25
Shoulder Radius	: 4.5
Elbow Radius	: 4.25
Wrist Radius	: 3.0
Distance bw 1st and 2-3 MTP	: 4

E. TRIAL RANDOMIZATION SHEET

MATLAB Array Editor: ans
Nov 7, 2008

Page 1
1:23:08 PM

	1			
1	13	–7.5	–3	+
2	14	–7.5	–4	
3	2	–5.5	–5	+
4	12	–7.5	–6	+
5	8	–6.5	–7	+
6	1	–5.5	–8	+
7	5	–5.5	–9	+
8	9	–6.5	–10	+
9	15	–7.5	–11	
10	11	–7.5	–12	
11	3	–5.5	–13	+
12	10	–6.5	–14	+
13	7	–6.5	–15	+
14	4	–5.5	–16	+
15	6	–6.5	–17	+

Force plate

z – 5000

xy – 1000

time constant off

S5-t1 – standing (all marker's)

S5-t2 – " (less marker)

F. MATLAB CODES

Random List Code

```
function out_list = rand_list(n)
n=15
orig_list = 1:n;
out_list = [];

while (~isempty(orig_list))
   num_left = length(orig_list);
   which_num = floor(num_left*rand(1)) + 1;

   out_list = [out_list orig_list(which_num)];
   if (which_num == 1)
      orig_list = [orig_list(which_num+1:num_left)];
   elseif (which_num == num_left)
      orig_list = [orig_list(1:which_num-1)];
   else
      orig_list = [orig_list(1:which_num-1) orig_list(which_num+1:num_left)];
   end
end

out_list = out_list'
```

Create Subject Data

```
clear all
close all

subjects(1).subnumber = 4;
subjects(1).triallist = [4 5 6 16 17];

subjects(2).subnumber = 5;
subjects(2).triallist = [5 7 8 9 10 13 14 16];

subjects(3).subnumber = 6;
subjects(3).triallist = [3 9 13 14 17];

subjects(4).subnumber = 7;
subjects(4).triallist = [ 5 6 9 10 16 17];

subjects(5).subnumber = 8;
subjects(5).triallist = [8 12 13 14 15 16 17];

subjects(6).subnumber = 9;
subjects(6).triallist = [4 5 6 7 10 12 13 14 16 17];

subjects(8).subnumber = 11;
subjects(8).triallist = [4 10 11 12 15];

subjects(9).subnumber = 12;
subjects(9).triallist = [4 9 10 12 14 15 17];

subjects(11).subnumber = 14;
subjects(11).triallist = [4 5 8 11 13 16 17];

subjects(12).subnumber = 15;
subjects(12).triallist = [3 5 9 11 13 16 18];

save subdata.mat subjects
```

Compute Angular Momentum

```
clear all
close all
start_dir = pwd;

RIGHT = 1;
LEFT = 2;

%% load structure with subject and trial information
load subdata.mat
nsubjects = length(subjects); % if =1, just do s4

%% for each subject
for sub_ind=1:nsubjects

%% go to directory with this subject's processed data directories
    subject_number = subjects(sub_ind).subnumber;
    subdir = ['C:\Users\BADE\Desktop\Tez Data\s' ...
        num2str(subject_number) '_DONE\c3d'];

%% for each of this subject's trials
    ntrials = length(subjects(sub_ind).triallist);
    for tri_ind = 1:ntrials
        outstr = ['Processing Subject #' num2str(subject_number) ...
            ', Trial ' num2str(tri_ind) ' of ' num2str(ntrials) '...'];
        disp(outstr)

%% define variables used in momentum calculations
%   nseg (scalar): number of segments
%   nframes (scalar): number of frames
%   seg_masses (nseg x 1): masses of segments
%   seg_MOIs (nseg x 3): principal moments of inertia for segments
%   seg_com_pos (nframes x nseg x 3): position of segment com in global
%   cop_pos (nframe x 3): position of center of pressure in global
%   com_velocity (nframes x nseg x 3): velocity of segment com in global
%   ang_velocity (nframes x nseg x 3): angular vel. of segment com in global
%   seg_orient (nframes x nseg x 9): rotation matrix between seg
%       and global s.t. v_global = R * v_segment
%   MTP_midpt (nframes x 3): location of the midpoint of the 1st MTHs
%   shoulder_pos (nframes x 3 x 2)
%   shoulder_moment (nframes x 3 x 2)
%   shoulder_frc (nframes x 3 x 2)
%   GRF (nframes x 3): GRF vector

%% go to directory with input data
        cd(subdir)
        dir_info = dir(pwd);
        for zz = 1:length(dir_info)
            if (~dir_info(zz).isdir)
                continue
            end
            trial_number = subjects(sub_ind).triallist(tri_ind);
            tr_str = num2str(trial_number);
```

```
        tr_str_len = length(tr_str);
        underscores_found = 0;
        for char_ind = 1:length(dir_info(zz).name)
            if (dir_info(zz).name(char_ind) == '_')
                underscores_found = underscores_found + 1;
            end
            if (underscores_found == 2)
                break
            end
            if (dir_info(zz).name(char_ind) == 't')
                t_loc = char_ind;
            end
        end
        if (underscores_found == 0)
            continue
        end
        if strcmp(dir_info(zz).name(t_loc+1:char_ind-1),tr_str)
            trialdir = [subdir '\' dir_info(zz).name];
            trial_lean_angle = eval(dir_info(zz).name(char_ind+1:char_ind+3));
            break
        end
    end
    cd(trialdir)

%% assign segment names
    segments = {'head';
        'thorax';
        'pelvis';
        'rt_upparm';
        'rt_forearm';
        'rt_hand';
        'lt_upparm';
        'lt_forearm';
        'lt_hand';
        'rt_thigh';
        'rt_shank';
        'rt_foot';
        'lt_thigh';
        'lt_shank';
        'lt_foot';};

%% define number of segments
    nseg = length(segments);

%% read in segment kinematics data exported from visual3D
    %  (filtered with 6 cutoff frequency)
    clear seg_com_pos ang_velocity com_velocity seg_orient
    for seg = 1:nseg

        seg_com_pos(:,seg,:)  = dlmread(['seg_com\' segments{seg} '.txt'],'', 0, 2);

        ang_velocity(:,seg,:) = dlmread(['ang_vel\' segments{seg} '.txt'],'', 0, 2);

        com_velocity(:,seg,:) = dlmread(['seg_vel\' segments{seg} '.txt'],'', 0, 2);
```

```
            seg_orient(:,seg,:) = dlmread(['seg_ang\' segments{seg} '.txt'],'', 0, 2);

        end

%% read in R and L shoulder forces, shoulder moments, and shoulder
%  joint positions

        clear shoulder_pos shoulder_moment shoulder_frc

        if (sub_ind == 1)
           olddir = pwd;
            shdir = ['C:\Users\BADE\Desktop\Tez Data\shoulder\s' num2str(subject_number) ...
                '\s' num2str(subject_number) '_t' tr_str];
            shoulder_pos(:,:,RIGHT)    = dlmread([shdir '\rt_shoulder.txt'],'', 0, 2);
            shoulder_pos(:,:,LEFT)     = dlmread([shdir '\lt_shoulder.txt'],'', 0, 2);
            shoulder_moment(:,:,RIGHT) = dlmread([shdir '\rt_sh_moment.txt'],'', 0, 2);
            shoulder_moment(:,:,LEFT)  = dlmread([shdir '\lt_sh_moment.txt'],'', 0, 2);
            shoulder_frc(:,:,RIGHT)    = dlmread([shdir '\rt_sh_force.txt'],'', 0, 2);
            shoulder_frc(:,:,LEFT)     = dlmread([shdir '\lt_sh_force.txt'],'', 0, 2);
        else
            olddir = pwd;
            shdir = ['C:\Users\BADE\Desktop\Tez Data\shoulder\s' num2str(subject_number) ...
                '\s' num2str(subject_number) '_t' tr_str];
            shoulder_pos(:,:,RIGHT)    = dlmread([shdir '\rt_shoulder.txt'],'', 0, 2);
            shoulder_pos(:,:,LEFT)     = dlmread([shdir '\lt_shoulder.txt'],'', 0, 2);
            shoulder_moment(:,:,RIGHT) = dlmread([shdir '\rt_sh_moment.txt'],'', 0, 2);
            shoulder_moment(:,:,LEFT)  = dlmread([shdir '\lt_sh_moment.txt'],'', 0, 2);
            shoulder_frc(:,:,RIGHT)    = dlmread([shdir '\rt_sh_force.txt'],'', 0, 2);
            shoulder_frc(:,:,LEFT)     = dlmread([shdir '\lt_sh_force.txt'],'', 0, 2);
            cd(olddir);
        end

%% determine nframes
        nframes = length(squeeze(seg_com_pos(:,1,1)));
        nframes2 = length(squeeze(ang_velocity(:,1,1)));
        nframes3 = length(squeeze(com_velocity(:,1,1)));
        nframes4 = length(squeeze(seg_orient(:,1,1)));

        if ((nframes ~= nframes2) || (nframes ~= nframes3) || (nframes2 ~= nframes3) || (nframes ~=
nframes4))
            disp('Kinematic frame number mismatch: skipping this trial!')
            continue
        end

%% read in segment inertial properties from Excel file
        clear seg_masses seg_MOIs MTP_midpt
        seg_masses = xlsread('seg_masses.xls', 'masses', 'b2:b16');

        seg_MOIs = xlsread('seg_masses.xls', 'MOI', 'b2:d16');

        MTP_midpt = dlmread('MTP_midpt2.txt','', 0, 2);
```

```
%% initialize big array for storage of H_toes and H_rsh and H_lsh
    H_toes = zeros(nframes,nseg,3);
    H_sho = zeros(nframes,nseg,3,2);

%% initialize array for strorage of WB COM
    WB_com_pos = zeros(nframes,3);

%% initialize array for storage of moment of GRF and weights about MTP midpoint
    MTP_grav_moment = zeros(nframes,nseg,3);
    MTP_grf_moment = zeros(nframes,3);
    WB_grav_moment = zeros(nframes,3);
    cop_pos = zeros(nframes,3);

%% read in Force Plate Data
    clear GRF cop_pos COM
    GRF = dlmread('GRF.txt','', 0, 0);

% COM = dlmread('COM.txt','', 0, 2);

% resample the data which is 1000 fps to 100 fps and convert into meters
    cop_pos = resample(dlmread('COP.txt','', 0, 0),100,1000)/1000;

%% main loop
    for frame=1:nframes

      temp = zeros(3,1);
      temp2 = zeros(3,1);
      temp3 = 0;
      for seg=1:nseg

% find r (3 x 1), difference between seg com and toes in global
        r_toes = zeros(3,1);
        r_toes = squeeze(seg_com_pos(frame,seg,:)) - MTP_midpt(frame,:)';

% same for shoulders
        r_sho = zeros(3,1,2);
        r_sho(:,:,RIGHT) = ...
          squeeze(seg_com_pos(frame,seg,:)) - shoulder_pos(frame,:,RIGHT)';
        r_sho(:,:,LEFT) = ...
          squeeze(seg_com_pos(frame,seg,:)) - shoulder_pos(frame,:,LEFT)';

% keep running total of mass * com position and mass
        temp2 = temp2 + squeeze(seg_com_pos(frame,seg,:)) * seg_masses(seg);
        temp3 = temp3 + seg_masses(seg);

% compute m (r x v)
        MV_cross_R_toes = zeros(3,1);
        MV_cross_R_toes = ...
          seg_masses(seg) * cross(r_toes,squeeze(com_velocity(frame,seg,:)));

% same for shoulders
        MV_cross_R_sho = zeros(3,1,2);
        MV_cross_R_sho(:,:,RIGHT) = ...
```

```
            seg_masses(seg) * cross(r_sho(:,:,RIGHT),squeeze(com_velocity(frame,seg,:)));
          MV_cross_R_sho(:,:,LEFT) = ...
            seg_masses(seg) * cross(r_sho(:,:,LEFT),squeeze(com_velocity(frame,seg,:)));

% transform seg angular velocity to seg frame
          R_gs = [squeeze(seg_orient(frame,seg,1:3))'; ...
                squeeze(seg_orient(frame,seg,4:6))'; ...
                squeeze(seg_orient(frame,seg,7:9))'];

          w_s = inv(R_gs) * squeeze(ang_velocity(frame,seg,:));

% compute angular momentum about segment COM in seg frame
          Iw = diag(squeeze(seg_MOIs(seg,:))) * w_s;

% compute H for this segment, transforming  Iw to ground frame
          H_toes(frame,seg,:) = MV_cross_R_toes + R_gs * Iw;

          if ((seg == 4) & (frame == 50))
            junk = 0;
          end

% same for shoulders
          H_sho(frame,seg,:,RIGHT) = MV_cross_R_sho(:,:,RIGHT) + R_gs * Iw;
          H_sho(frame,seg,:,LEFT)  = MV_cross_R_sho(:,:,LEFT) + R_gs * Iw;

% compute moment of each segment weight about MTP midpoint
          this_seg_wt = [0 -seg_masses(seg)*9.81 0];
          MTP_grav_moment(frame,seg,:) = cross(r_toes, this_seg_wt);
          temp = WB_grav_moment(frame,:) + squeeze(MTP_grav_moment(frame,seg,:))';

% compute Linear momentum for each segment
          Linear_momentum(frame,seg,3) = seg_masses(seg) *  com_velocity(seg,3);

        end
% seg

% compute moment of GRF about MTP midpoint
        mtp_to_cop = cop_pos(frame,:) - MTP_midpt(frame,:);

        MTP_grf_moment(frame,:) = cross(mtp_to_cop, GRF(frame,:));
        WB_grav_moment(frame,:) = temp;
        WB_com_pos(frame,:) = temp2/temp3;
        WB_com_pos(1,3)=WB_com_pos(2,3);

    end % frame

%% create time vector for plotting
    samp_freq = 100;
    delta_t = 1/samp_freq;
    time = 0:delta_t:(nframes-1)*delta_t;

%% Store information about this trial in output structure
    outdata(sub_ind,tri_ind).H = H_toes;
```

```matlab
        outdata(sub_ind,tri_ind).subject_number = subject_number;
        outdata(sub_ind,tri_ind).trial_number = trial_number;
        outdata(sub_ind,tri_ind).time = time;
        outdata(sub_ind,tri_ind).trial_lean_angle = trial_lean_angle;
        trial_counts(sub_ind) = ntrials;
        outdata(sub_ind,tri_ind).WB_com_pos = WB_com_pos;
        outdata(sub_ind,tri_ind).cop_pos = cop_pos;
        outdata(sub_ind,tri_ind).MTP_midpt = MTP_midpt;
        outdata(sub_ind,tri_ind).GRF = GRF;

%% Compute Whole Body (without arms) Angular Momentum
        outdata(sub_ind,tri_ind).H_not_arms = zeros(nframes,3);
        for coor = 1:3
            outdata(sub_ind,tri_ind).H_not_arms(:,coor) = (H_toes(:,1,coor)+H_toes(:,2,coor)+
H_toes(:,3,coor)+...
                H_toes(:,10,coor)+H_toes(:,11,coor)+H_toes(:,12,coor)+...
                H_toes(:,13,coor)+H_toes(:,14,coor)+H_toes(:,15,coor));
        end

%% Compute Angular Momentum of Arms about Toes
        outdata(sub_ind,tri_ind).H_arms = zeros(nframes,3);
        for coor = 1:3
            outdata(sub_ind,tri_ind).H_arms(:,coor) = (H_toes(:,4,coor)+H_toes(:,5,coor)+
H_toes(:,6,coor)+...
                H_toes(:,7,coor)+H_toes(:,8,coor)+H_toes(:,9,coor));
        end

%% Compute Angular Momentum of Arms about Shoulders
        outdata(sub_ind,tri_ind).H_arms_sho = zeros(nframes,3,2);
        for coor = 1:3
            outdata(sub_ind,tri_ind).H_arms_sho(:,coor,RIGHT) = H_sho(:,4,coor,RIGHT)+...
                H_sho(:,5,coor,RIGHT)+H_sho(:,6,coor,RIGHT);
            outdata(sub_ind,tri_ind).H_arms_sho(:,coor,LEFT)  = H_sho(:,7,coor,LEFT)+...
                H_sho(:,8,coor,LEFT)+H_sho(:,9,coor,LEFT);
        end

    end % trials loop
end % subjects loop

%% save processed data to file

cd(start_dir)
save falldata_sh.mat outdata nsubjects segments trial_counts
```

Extract Data for Statistical Analysis

```
clear all
close all

RIGHT = 1;
LEFT = 2;

%% load processed momentum data

load falldata_sh.mat

masses = [76.4 67.3 91.4 117.0 62.0 52.0 110.0 71.0 56.0 58.0 76.0 60.0];

masses65 = [76.4 67.3 91.4 117.0 62.0 52.0 71.0 56.0 76.0 60.0];

recovery = [4 2 5 5 4 1 5 3 4 4 5 5 4 5 2 5 4 1 5 2 5 2]';

recovery55 = [4 5 4 5 4 5 4 5 5 1 5 5]';
recovery65 = [2 5 1 3 4 5 2 4 2 2]';

anova_table = [];
anova_table2 = [];

%% for each subject
for sub_ind = 1:nsubjects

    temp55 = [];
    temp65 = [];

%% for each trial
    for tri_ind = 1:trial_counts(sub_ind)

        H_arms_sho = outdata(sub_ind,tri_ind).H_arms_sho;

        sub = outdata(sub_ind,tri_ind).subject_number;
        tri = outdata(sub_ind,tri_ind).trial_number;
        lean = outdata(sub_ind,tri_ind).trial_lean_angle;
        GRF_max  = max(outdata(sub_ind,tri_ind).GRF(:,2));

        if outdata(sub_ind,tri_ind).WB_com_pos(length(outdata(sub_ind,tri_ind).WB_com_pos(:,3)),3)
> 0

        SS_t = find(outdata(sub_ind,tri_ind).WB_com_pos(:,3) >
outdata(sub_ind,tri_ind).WB_com_pos(length(outdata(sub_ind,tri_ind).WB_com_pos(:,3)),3)*0.95 ...
            & outdata(sub_ind,tri_ind).WB_com_pos(:,3) <
outdata(sub_ind,tri_ind).WB_com_pos(length(outdata(sub_ind,tri_ind).WB_com_pos(:,3)),3)*1.05)*
0.01;
        SS_t = SS_t(1);

        else
```

```
            SS_t = find(outdata(sub_ind,tri_ind).WB_com_pos(:,3) <
outdata(sub_ind,tri_ind).WB_com_pos(length(outdata(sub_ind,tri_ind).WB_com_pos(:,3)),3)*0.95 ...
            & outdata(sub_ind,tri_ind).WB_com_pos(:,3) >
outdata(sub_ind,tri_ind).WB_com_pos(length(outdata(sub_ind,tri_ind).WB_com_pos(:,3)),3)*1.05)*
0.01;
        SS_t = SS_t(1);
        end

        if ((sub == 12) & (tri == 8))
            continue;
        end

        frame_num = length(squeeze(H_arms_sho(:,1,RIGHT)));
%         if (frame_num > 150)
%             frame_num = 150;
%         end

        [min_H_R,min_H_R_i] = min(squeeze(H_arms_sho(1:frame_num,1,RIGHT)));
        min_Hn_R = min_H_R/masses(sub_ind);
        min_H_R_t=min_H_R_i*0.01;
        [max_H_R,max_H_R_i] = max(squeeze(H_arms_sho(1:frame_num,1,RIGHT)));
        max_Hn_R = max_H_R/masses(sub_ind);
        max_H_R_t=max_H_R_i*0.01;

        [min_H_L,min_H_L_i] = min(squeeze(H_arms_sho(1:frame_num,1,LEFT)));
        min_Hn_L = min_H_L/masses(sub_ind);
        min_H_L_t=min_H_L_i*0.01;
        [max_H_L,max_H_L_i] = max(squeeze(H_arms_sho(1:frame_num,1,LEFT)));
        max_Hn_L = max_H_L/masses(sub_ind);
        max_H_L_t=max_H_L_i*0.01;


 %% Anova Table herbir denek'in tum triallari icin yukarida cekilen
 %% degerlerinin tablosu
        anova_table = [anova_table; sub tri lean min_H_R min_Hn_R max_H_R max_Hn_R ...
            min_H_L min_Hn_L max_H_L max_Hn_L min_H_R_t max_H_R_t min_H_L_t max_H_L_t
GRF_max SS_t];

        if (lean == 5.5)
            temp55 = [temp55; min_H_R min_Hn_R max_H_R max_Hn_R ...
                min_H_L min_Hn_L max_H_L max_Hn_L min_H_R_t max_H_R_t min_H_L_t
max_H_L_t GRF_max SS_t];
        else
            temp65 = [temp65; min_H_R min_Hn_R max_H_R max_Hn_R ...
                min_H_L min_Hn_L max_H_L max_Hn_L min_H_R_t max_H_R_t min_H_L_t
max_H_L_t GRF_max SS_t];
        end
    end
%% Anova Table2 ise herbir denek'in tum triallarinin 5,5 ve 6,5
%% derecelerdeki ortalamalarinin tablosu
    if (~isempty(temp55))
        anova_table2 = [anova_table2; sub 5.5 mean(temp55,1)];
    end
```

```
    if (~isempty(temp65))
        anova_table2 = [anova_table2; sub 6.5 mean(temp65,1)];
    end

end

anova_table2 = [anova_table2 recovery];

anova_table55= anova_table2(1,:);
anova_table65= anova_table2(2,:);

for i=3:length(anova_table2)

if anova_table2(i,2)==5.5
    anova_table55 = [anova_table2(i,:);anova_table55];
else
    anova_table65 = [anova_table2(i,:);anova_table65];
end
end

anova_table55= flipdim(anova_table55,1);
anova_table65= flipdim(anova_table65,1);

Max_H_arms_55 = anova_table55(:,5);
Max_H_arms_65 = anova_table65(:,5);

save for_anova.txt anova_table -ascii
save for_anova2.txt anova_table2 -ascii
```

G. CURRICULUM VITAE

Name:	Emre Ak
Address :	Physical Education and Sports Dep., no:409 Faculty of Education, Middle East Technical University 06531 Çankaya
Telephone	+90 312 210 3663 (office) +90 533 167 3194 (mobile)
E-mail	emreak@gmail.com
Birth Place and Year:	Düzce, 02.07.1977
Marital Status:	Married – 1 daughter
Current Occupation:	Research Assistant

Research interests:

Sports Biomechanics
Training Theory
Fitness Assessment

Education:

PhD (ongoing),	Middle East Technical University Department of Physical Education and Sports, Ankar Turkey
Master of Science (2004),	Kırıkkale University, Faculty of Education, Department of Physical Education and Sports, Kırıkkale, Turkey.
Bachelor's Degree (2000),	Abant Izzet Baysal University, School of Physical Education and Sports, Department of Physical Education Teacher Education, Bolu, Turkey.

Academic Positions:

Research Assistant, Middle East Technical University Department of Physical Education and Sports, Ankara, Turkey	2005 ~
Visiting Scholar, Pennsylvania State University, School of Health and Human Development, Department of Kinesiology, Biomechanics Lab	2007 - 2009
Research Assistant, Faculty of Education, Department of Physical Education and Sports	2001 - 2005

Professional Responsibilities:

International Table Tennis Federation, Science Committee Full Member	2009 ~
Turkish Table Tennis Federation Board of Education Member	2006 ~

Language Skills:	KPDS	ÜDS
English	84	82.5
Turkish	Native	

PUBLICATIONS

International

1. Celik O., Salci Y., **Ak E.**, Kalaci A., Korkusuz F., (2012), Serum cartilage oligomeric matrix protein accumulation decreases significantly after 12weeks of running but not swimming and cycling training - A randomized controlled trial, The Knee, Jul 5. [in press].
2. Mutlu Cuğ, **Emre Ak,** Recep Ali Özdemir, Feza Korkusuz and David G. Behm, (2012), The effect of instability training on knee joint proprioception and core strength, Journal of Sports Science and Medicine, September 2012 - Volume 11, Issue 3, 468 – 474.

3. **Ak, E.,** Koçak, M.S., (2010). Coincidence-Anticipation Timing and Reaction Time in Youth Tennis And Table Tennis Players, Perceptual and Motor Skills, 110, 3, 879-887.
4. Uygur, M., Goktepe, A., **Ak, E.,** Karabork, H., Korkusuz, F., (2010), The Effect of Fatigue on the Kinematics of Free Throw Shooting in Basketball, Journal of Human Kinetics, 24, 51-56.
5. Goktepe, A., **Ak, E.,** Karabörk, H., Sogut, M., Korkusuz, F., (2009). Joint angles during successful and unsuccessful tennis serves kinematics of tennis serve, Joint Diseases & Related Surgery, Vol 20, no:3.
6. **Ak, E.**, Goktepe, A., Karaborg, H., Cicek, S., Korkusuz, F., (2007). Photogrammetric Analysis of Penalty Kick in Soccer, Journal of Sports Science and Medicine, Supplementum 10, 01 February 2007.

National

1. Söğüt, M., **Ak, E.** & Koçak, S. (2009) Coincidence timing accuracy of junior tennis players. Spor Bilimleri Dergisi. Spor Bilimleri Dergisi, Hacettepe J. of Sport Sciences, 2009, 20 (1), 1–5.

Book

1. **Ak, E.**, Korkusuz, F., Yılmaz, L., Salcı, Y., (2007). Sık Karşılaşılan Spor yaralanmaları ve İlk Müdahale. ODTÜ Yayıncılık, Ankara.

Book Chapters

1. **Ak, E**, Yıldırım A, Çiçek Ş, Korkusuz F. Comparison of physiological profiles of soccer players in U17, U19, U21 and over-21 age groups. In. Science and Football VI, Eds. Thomas Reilly and Feza_Korkusuz, Routledge, London, 2009, pp. 360-3.
2. Yıldırım A, **Ak, E**, Korkusuz F, Çiçek Ş. Physiological profiles of soccer players with respect to playing positions. In. Science and Football VI, Eds. Thomas Reilly and Feza Korkusuz, Routledge, London, 2009, pp. 370-3.

CONFERENCE PAPER

1. **Ak, E.,** Çimen, O., (2011, May). The Effect of Split Step on Kinetic and Temporal Variables in Table Tennis, 12^{nd} ITTF Sport Science Congress, Rotterdam, Netherlands.
2. **Ak, E.,** Piazza, S.J., (2010, September). Modulation of the Angular Momentum of the Arms
   in Fall Recovery, 1^{st} International Congress on Applied Bionics and Biomechanics, Venice, Italy.

3. **Ak, E.,** Piazza, S.J., (2009, October). Methodology of Investigating the Use of the Arms in Fall Recovery, 6th European Sports Medicine Congress, Antalya, Turkey.
4. **Ak, E.,** Goktepe, A., Karaborg, H., Cicek, S., Korkusuz, F., (2007, January). Photogrammetric Analysis Of Penalty Kick in Soccer, 6th World Congress on Science & Football, Antalya, Turkey.
5. Goktepe, A., **Ak, E.,** Karaborg, H., Cicek, S., Korkusuz, F., (2008, March). Kinematic analysis of penalty kick in soccer, 6th International Congress on Physical Education and Sport Science Iran, Kish.
6. Uygur, M., Goktepe, A., **Ak, E**., Karaborg, H., Korkusuz, F., (2008, March). Fatigue effects free throw shooting in basketball, 6th International Congress on Physical Education and Sport Science Iran, Kish.
7. **Ak, E.,** (2007, May). The use of split step by table tennis players in Turkey, 10th ITTF Sports Sciences Congress, Zagreb, Croatia.
8. **Ak, E.,** (2007, May). The Effect of Split Step on Serve Return Performance in Table Tennis, 10th ITTF Sports Sciences Congress, Zagreb, Croatia.
9. Erol, A.E., Çimen O., Ayan, V., Mülazımoğlu, O., **Ak, E.,** (2007, May). The determination of anthropometric characteristics of Turkish children tends to be table tennis players, 10th ITTF Sports Sciences Congress, Zagreb, Croatia.
10. Söğüt, M., Koçak, S. & **Ak, E.** The effects of age and gender on coincidence timing accuracy of junior tennis players. In proceedings of the 4th International Mediterranean Sports Science Congress, 9-11 November 2007. Antalya, Turkey.
11. **Ak, E.**, Söğüt, M., Uygur, M., Yıldırım, A., Cuğ, M., Korkusuz, F. & Özgider, C. The effects of 6-week callisthenic training on serve speed of 10–13 years old tennis players. In proceedings of the 4th International Mediterranean Sports Science Congress, 9-
12. **Ak, E.,** Yıldırım, A., Cicek, S., Korkusuz, F., (2007, January). Physiological profiles of soccer players: U17, U19, U21 and over21, 6th World Congress on Science & Football, Antalya, Turkey.
13. Yıldırım, A., **Ak, E.,** Cicek, S., Korkusuz, F., (2007, January). Physiological profiles of soccer players with respect to playing positions, 6th World Congress on Science & Football, Antalya, Turkey.
14. **Ak, E.,** Kocak, S., (2006, November). Comparison of Reaction Time and Coincident-timing Accuracy of Tennis and Table Tennis Players, (2006, November) 9th International Sport Sciences Congress, Mugla, Turkey.

H. TÜRKÇE ÖZET

DENGE KAYBI SIRASINDA DENGENIN YENIDEN KAZANILMASI ICIN KOLLARIN KULLANILMASININ ARASTIRILMASI

Giriş

Dünya sağlık örgütü verilerine göre, düşmelere bağlı sakatlıklar ve ölümler kazalara bağlı sakatlıklar arasında trafik kazalarının ardından ikinci sırada yer almaktadır (WHO, 2012). Dünya üzerinde kazalar sonucunda ölen insanların sayısı 424.000 olarak rapor edilmiştir ve bu kazaların büyük çoğunluğu orta veya düşük gelirli ülkelerde olmaktadır.

Düşmeler nedeniyle ortaya çıkan finansal boyut da bir o kadar yüksektir. Düşmeler sonucunda Kanada'da kullanılan yıllık bütçe, yaklaşık 6 milyon dolar olarak hesaplanmıştır. Bu nedenle Amerika, Kanada, Avustralya ve İngiltere gibi gelişmiş ülkelerde düşmeler, düşmelerin önlenmesi için alınması gereken önlemler ve toparlanma stratejileri üzerine araştırmalar yapılmaktadır ve bu çalışmalara destek verilmektedir.

Düşme, bir kişinin denge kaybına veya başka bir dış etkene bağlı olarak yere veya daha alt seviyeye inmesi olarak tanımlanmaktadır. Ayakta durabilmek için vücut, çok karmaşık duyusal ve merkezi sinir sistemini ile geri bildirim sistemi kullanmaktadır (Eurich & Milton, 1996). Dengeyi korumak için kullanılan kasların ne kadar kasılacağı bilgisi kasların içerisinde bulunan kas iğcikleri, golgi tendon organı ve ayak tabanında bulunan sensörler tarafından sağlanır (Moss & Milton, 2003).

Düşmeler, vestibüler, somotosensör ve görme bozuklukları, yaşlanmaya bağlı merkezi sinir sistemi problemleri, hareket sistemindeki kısıtlamalara bağlı oluşabilir (Horak et al., 1989).

Düşmeler arasında en sık kayma ve yerde bulunan bir engele takılma sonucu oluşan düşmeler meydana gelir ve bunlar tüm düşmeler arasında yaklaşık % 30-50 yi oluşturur (Cumming & Klineberg, 1994; Gabell et al., 1985; Topper et al., 1993).

Düşmeleri anlamak için farklı türde deneysel çalışmalar yapılmıştır. Bunlar, laboratuar ortamında yapılan, yürüme yolunda aniden ortaya çıkan bir engel (Pavol et al., 1999; M. Pijnappels et al., 2004), kaygan zemin (Cham & Redfern, 2001; Troy & Grabiner, 2006) veya yürüyüş sırasında adım salınma safhasında ayak bileğine bağlanan bir ip vasıtasıyla adımın engellenmesi şeklindedir (Forner Cordero et al., 2003; Smeesters et al., 2001). Ayakta dururken ise bele bağlanan bir ipin aniden çelikmesi (Luchies et al., 1994; Rogers et al., 2001), üzerinde durulan platformun aniden belli bir yöne doğru eğilmesi (McIlroy & Maki, 1996; Pavol et al., 2002) veya duvara bağlanan bir halat yardımıyla bireyin belli bir açıda öne, geriye veya yanlara doğru eğilmesi şeklinde yapılmıştır. Bu çalışmada, en son bahsedilen, katılımcının bir halat yardımıyla belli açılarda öne doğru eğilmesi ve aniden serbest bırakılması yöntemi kullanılmıştır.

Ayakta dengede durabilmek için merkezi sinir sistemi görsel bilgileri kullanır. Vestibüler ve somotosensör girdiler uygun motor uyaranları oluşturarak ağırlık merkezinin destek alanı içerisinde kalmasını sağlar (Dawn Crystal Mackey, 2004). Denge kaybı sırasında dengede kalabilmek amacıyla yapılan hareketler iki kategoride incelenebilir. Bunlar ayaklar sabit ve ayaklar hareketli stratejilerdir. Ayaklar sabit olarak kullanılan stratejiler daha çok düşük şiddetli denge kayıplarında kullanılır ve otonomik bir strateji olarak değerlendirilir. Ayak bilekleri, bacaklar ve kalça kaslarının kasılması veya kolların kullanılması şeklinde görülebilir. Kullanılan ikinci stratejide ise bir veya birden çok adım alma veya çevredeki herhangi bir sabit yeri tutma şeklinde olabilir. Denge merkezinin değiştiği bu strateji genellikle şiddeti yüksek olan denge kayıplarında kullanılır.

Vücudumuzda iki çeşit hareket kontrol sistemi bulunmaktadır. Bunlar açık döngülü ve kapalı döngülü sistemlerdir. Kapalı döngü kontrol sistemi kas boyundaki uzamaları tespit ederek sakatlığı önlemek amacıyla bu bilgiyi omurilikteki sinir merkezlerine ulaştıran kas iğcikleri ve kasların tendonlara bağlantı noktalarında olan ve kasın gerilme şiddetini ayarlayan golgi tendon organından gelen uyarılara bağlıdır. Kapalı döngü kontrol sistemi, beynin motor korteks bölgesi tarafından kontrol edilir ve uyaranların kaslarımıza iletilebilmesi için yaklaşık 400*600 milisaniye gibi bir süreye ihtiyaç vardır. Diğer yandan açık döngü kontrol sistemi beynin arka tarafında bulunan beyincik tarafından kontrol edilir ve ani balistik hareketlerden sorumludur. Hareketler otomatik bir şekilde gerçekleşir.

Denge kaybı sırasında toparlanabilmek için kasların kullanımı incelendiğinde yaşlı bireylerin kaslarını gereğinden fazla aktive ettikleri, genç bireylerin ise toparlanabilmek için yeterli olan miktarda kullandıkları bulunmuştur (Manchester et al., 1989).

Denge kaybı beklenmedik ve aniden oluşan bir olaydır. Acaba dengemizi yeniden kazanmak düşünerek hareketlerimizi tasarlamak için yeteri kadar vaktimiz var mıdır? Eğer dengemizi yeniden koruyabilmek için gerekli olan refleksif hareketleri uygulayabilirsek bir sonraki aşamadan uygun hareketleri yaparak düşmeden dengemizi koruyabilme şansımız olabilir. Bu refleksif hareketlerin uygun egzersizler yapılarak geliştirilebildiği bilimsel çalışmalar ile gösterilmiştir.

Denge örneğinden yola çıkacak olursak, sportif aktivitelerin hemen hemen hepsinde dengenin performans açısından ne kadar önemli olduğunu vurgulamaya gerek yoktur. Cimnastik, kayak ve diğer tüm artistik spor branşlarında denge birincil beceri olarak öne çıkarken diğer spor branşlarında tekniği düzgün ve etkili uygulayabilmek için bir ön şart olduğu söylenebilir.

Bu çalışmadan dengenin kazanılmasında açısal momentumun ve kol çevirme hızının etkisi incelenmiştir. Açısal momentum bir çok faktörün bir araya gelmesi ile hesaplanan bir olgudur. Bu çalışmada farklı denge kaybı şiddetleri uygulandığında bireylerin açısal momentumlarında herhangi bir farklılık olup olmadığı incelenmiştir. Buradan çıkartılacak olan sonuç karmaşık hareketler içeren spor dallarında teknik uygulamalar sırasında vücut uzuvlarının kullanımı ile ilgili önemli bilgiler vereceği düşünülmektedir.

Bu çalışmanın sonunda denge kaybı şiddetinin

- maksimum pozitif ve negatif açısal momentuma

- maksimum pozitif ve negatif kol açısal hızına

- yere uygulanan maksimum kuvete

- toparlanma süresine

- omuzda oluşan torka olan etkilerinin nasıl olacağı sorusu yanıtlanmaya çalışılmıştır.

Materyal ve Metod

Bu çalışmaya Pennsylvania State Üniversitesinde okuyan, altı erkek ve 4 kadın katılımcı kendi istekleri ile katılmışlardır (Yaş: 24.0 ± 2.0 yıl; Boy: 1.73 ± 0.1 m; Vücut Ağırlığı: 74.8 ± 21.1 kg). Katılımcılara deney ile ilgili tüm detaylar ve riskler anlatıldıktan sonra Pennsylvania State Üniversitesi, Etik Komitesi tarafından onaylanan Aydınlatılmış Onam Formunu imzalamaları istenmiştir. Katılımcıların herhangi bir kas-hareket sistemi rahatsızlığı olmaması ön şart olarak belirtilmiştir.

Katılımcıların deney sırasında tayt veya vücutlarını saran kıyafetler giymeleri istenmiştir. Bu hareket yakalama sistemi işaretleyicilerinin hareket etmemesi açısından önemlidir. Deneyler ayakkabının olası olumlu veya olumsuz etkilerini önlemek amacıyla çıplak ayakla yapılmıştır. Deney sırasında olası yorgunluğu önlemek amacıyla her denemenin ardından 30 sn ve her beş denemenin ardından 3 dakika dinlenme arası verilmiştir.

Katılımcı ölçüm için laboratuara geldiğinde yukarıda bahsedilen prosedürler tamamlandıktan sonra 72 adet yansıtıcı işaretleyici Appendix C'de gösterildiği gibi katılımcının üzerine yerleştirilmiştir. Bu işaretleme yöntemi c-motion tarafından önerilmiştir. İşaretleyicilerin yerleştirilmesinin ardından katılımcının kuvvet platformu üzerinde ayaklar omuz genişliğinde kollar dirseklerden yaklaşık 90 derece bükülü bir şekilde sabit durması istenmiştir. Bu pozisyon bir sonraki aşamada model oluşturmak amacıyla kullanılacak olan sabit duruş pozisyonudur. Yaklaşık 2 saniyelik bir kayıt yapıldıktan sonra katılımcının üzerinden yalnızca model oluşturmak için kullanılacak ve genelde eklemler üzerine yerleştirilmiş olan 18 adet işaretleyici çıkartılmıştır. Kalan 54 işaretleyici katılımcıların hareketlerini takip edebilmek için yeterli olan işaretleyicilerdir.

Tüm ölçümler Pennsylvania State Üniveristesi, Kinesioloji Departmanı, Biyomekanik Laboratuarında yapılmıştır. Laboratur içerisinde bulunan yürüme analizi için kullanılan zemine sabitlenmiş iki adet 60x40cm'lik Kistler kuvvet platformu (model 9286AA) bulunmaktadır. Ölçümler sırasında katılımcıların bu platformlardan bir tanesinin üzerinde durmaları istenmiştir. Kuvvet platformu öne geriye, sağa,sola ve dikey yönde kuvvetleri tespit edebilme özelliğine sahiptir.

Laboratuarda aynı zamanda altı kameralı Eagle Digital Kamera Sistemi (Motion Analysis Coorporation) bulunmaktadır. Herbir kameranın 1280x1024 boyutlarında 1.3 milyon piksel çözünürlükte ve 100 kare yakalama hızı özelliği vardır. Katılımcıların üzerlerine yerleştirilen işaretleyiciler sistem tarafından tespit edilerek üç boyutlu pozisyon verisi olarak kaydedilir. Hareket yakalama sisteminin yazılımı olan EvaRT yazılımı aynı zamanda işaretleyicilerin pozisyonlarında oluşan boşlukları da doldurma özelliği bulunmaktadır. Boşluklar doldurulurken segmentler üzerine yerleştirilen cluster (birbirlerinden bağımsız hareket edemeyen) işaretleyici setleri kullanılmıştır.

İçaretleyicilerin üç boyutlu pozisyonlarının kaydedilmesinin ardından C-Motion firmasının Visual 3D yazılımı kullanılarak herbir katılımcılara özel iskelet modeller oluşturulmuştur. Visual 3D programı aynı zamanda analizler için gerekli olan tüm kinetik ve kinematik verlerin hesaplanması için kullanılmıştır.

Modellerin oluşturulması ve gerekli verilerin kaydedilmesinin ardından tüm hesaplamalar için yazılmış olan MATLAB kodu kullanılmıştır.

İstatistiksel analizler IBM SPSS 18 kullanılarak yapılmıştır. Katılımcı sayısından dolayı analizler Wilcoxon Summed Ranked Test kullanılarak yapılmıştır. Değişkenlerin birbirleri ile olan ilişkilerinin incelemek amacıyla Linear Regression analizi yapılmıştır.

Daha sonra katılımcıların ölçümleri kendileri için rastgele bir şekilde oluşturulmuş olan deney açısı sırasıyla yapılmıştır. Bu rastgele ölçüm sırası MATLAB kodu kullanılarak oluşturulmuştur. Katılımcıların 5.5, 6.5 ve 7.5 derecelerde öne doğru eğik bir şekilde durmaları bellerinden arka taraftaki duvara bağlanan bir halat sayesinde olmuştur. Bu halatın orta kısmında bir elektromagnet bulunmaktadır. Bu eketromagnet halatın her iki kısmının birleşmesini sağlamaktedır. Elektromagnet serbest bırakıldığında ise aradaki bağ koparak katılımcıların öne doğru düşmeleri sağlanmıştır. 5.5, 6.5 ve 7.5 derecelerde öne doğru eğim açısını hesaplamada basit trigonometrik işlemler kullanılmıştır. Açının miktarını belirleyecek olan halatın uzunluğu, ayak bileği eklemi ile duvar arasındaki mesafe, duvardaki halatın bağlanma kancasının yüksekliği, ayak bileği eklemi ve duvardaki çengelin arasındaki mesafe ve akak bileği ve halatın kalçaya bağlanma noktasının yüksekliği kullanılarak hesaplanmıştır (şekil 8). Katılımcıların öne doğru eğilme açıları aynı zamanda her ölçümün ardından ayak bileği eklemi üzerindeki işaretleyici, kalça eklemi üzerindeki işaretleyici ve omuz üzerindeki işaretleyicilerin 3 boyutlu pozisyonları kullanılarak hesaplanmıştır. Eğer 0.2 derecenin üzerinde bir açı hatası varsa o denene analiz için kullanılmamıştır. Katılımcılar yukarıda belirtilen açılarda öne doğru eğilirken vücutlarının düz bir hat üzerinde olması istenmiştir ve denge kaybı yaşadıkları sırada adım almadan önce toparlanabilmek için yapabildikleri herşeyi yapmaları istenmiştir. Katılımcı ve prosedür hazır olduğunda ise katılımcılara “hazır” komutu verilmesinin ardından 5 saniye içerisinde elektromagnet devre dışı bırakılmıştır. Her testin ardından bir sonraki açı için halatın uzunluğu değiştirilmiştir. Katılımcıların düşmelerini engellemek için tavana bağlı bir güvenlik yeleği giydirilmişir.

Katılımcıların hiçbirisi 7.5 dereceden toparlanamamıştır. Bu da Mackey ve Robinovic (2006)’in bulduğu toparlanılabilen maksimum açının 7.2 bulgusunu desteklemiştir.

Deney sırasında izlenilen yol ve yaklaşık süreleri aşağıdaki tabloda verilmiştir.

Aşamalar	Süre
1- Test prosedürünin katılımcıya açıklanması	5 dk.
2- Bilgilendirilmiş onam formunun okunması ve imzalanması	3 dk.
3- İşaretleyicilerin yerleştirilmesi	30 dk
4- Açı ayarı için halat uzunluklarının belirlenmesi	10 dk.
5- Ölçümlerin yapılması ve verilerin control edilmesi	20 dk.
6- İşaretleyicilerin çıkartılması	10 dk.
7- Anthropometrik ölçümler	5 dk.

Tüm katılımcılardan veri toplandıktan sonra işaretleyilerin pozisyon verilerindeki boşluklar hareket yakalama sisteminin yazılımı EvaRT ile doldurulmuştur ve 6 hz ile butterworth eleme uygulanmıştır. Daha sonra datalar c3d fomatına çevirilerek kaydedilmiş ve Visual 3D (c-motion) programı ile tüm katılımcılar için özel iskelet modeller oluşturulmuştur. Model oluşturmada katılımcıların vücutları üzerine yerleştirler işaretleyiciler kullanılmıştır. Aynı zamanda antropometrik ölçümler de kullanılarak her katılımcıya ait vücut ölçüleri model üzerine uygulanmıştır. Daha sonra testler esnasında kollarda ve tüm vücutta oluşan açısal momentumları hesaplamak için baş, kollar, gövde, pelvis, üst bacak, alt bacak, ayaklar, üst kollar ve ön kollar olmak üzere toplam 15 vücut parçasına ait ağırlık merkezi pozisyonları, ağırlık merkezi açısal hızları, ağırlık merkezi linear hızları, açıları, tüm vücut ağırlık merkezi pozisyonu ve yere uygulanan dikey kuvvet miktarı (Fz) herbir katılımcı için hesaplanarak katılımcılara ait dosyalara kaydedilmiştir. Verilerin kaydedilmesinin ardından yazılan MATLAB kodu ile kollar ve tüm vücut için açısal momentum, ön kolların açısal hızları, toparlanma zamanı ve omuz torkları hesaplanmıştır.

Sonuçlar

Yapılan analizler sonucunda katılımcıların hiçbirisinin 7.5 dereceden toparlanamadıkları görülmüştür. 5.5 ve 6.5 derecelerden toparlanma sayıları incelendirğinde 5.5 derecede ortalama 4.7 (kez/5) ve 6.5 dereceden 3 (kez/5) olarak bulunmuştur ve bu fark istatistiksel olarak anlamlıdır ($p=0,005$).

Açısal momentum skorları iki farklı şekilde incelenmiştir. Birincisi omuz eklemi baz alınarak açısal momentumun hesaplanması şeklindedir. Bu bize kolların geri kalan vücut uzuvlarına bakılmaksızın dengenin korunması için ne kadar etkili olduğunu göstermektedir. İkincisi ise ayak parmak uçlarının orta noktası baz alınarak hesaplanan kolların açısal momentum değerleri. Bu da bize ayaktan başlayarak tüm vücut uzuvlarının etkilerini göze alarak kolların açısal momentumunu vermektedir.

Wilcoxon Signed Ranked Test sonucunda sağ ve sol kollar omuz eklemine gore hesaplanan minimum negatif açısal momentum sonuçları 5.5 ve 6.5 dereceler için karşılaştırılmıştır. Sonuçlar göstermiştir ki hem sağ hem de sol kol açısal momentumları 5.5 ve 6.5 arasında anlamlı farklılıklar vardır ($p=0.005$). Bu farkın yaklaşık % 50 oranında olduğu görülmüştür. Açısal momentum değerleri katılımcıların vücut ağırlıkları kullanılarak normalize edildiğinde de her iki kol için 5.5 ve 6.5 dereceler arasında anlamlı farklar bulunmuştur ($p=0.005$).

Maksimum negatif açısal momentum değerleri kolların vücudun önünde aşağıya doğru çevirilmesi sırasında oluşan momentumu göstermektedir. Kolların vücudun arkasında yukarıya doğru çevirilmesinde ise positif açısal momentum oluşmaktadır (koordinat sisteminden dolayı).

Positif açısal momentum değerleri incelendiğinde ise sağ ve sol kollar için 5.5 ve 6.5 dereceler arasında herhangi bir anlamlı farklılık bulunmamıştır.

Kollarda oluşan açısal momentumları ayak parmak uçlarının orta noktası baz alınarak hesaplandığında ise 5.5 ($p=0,005$) ve 6.5 ($p=0,047$) dereceler arasında hem pozitif hem de negatif açısal momentumlar arasında istatistiksel olarak anlamlı farklılıklar bulunmuştur.

Kolların çevirilme hızları incelendiğinde ise 6.5 derecede (10.503 rad/sn) 5.5 dereceye (6.709 rad/sn) oranla anlamlı olarak daha hızlı çevirildiği bulunmuştur ($p=0,005$).

5.5 ve 6.5 dereceler için yere dikey uygulanan kuvvet, toparlanma zamanları, sağ ve sol omuz torkları incelendiğinde ise herhangi bir anlamlı farklılık bulunamamıştır.

Kolların çevirilme hızının vücut ağırlığı, boy, kolların eylemsizlik momenti ve vücut kitle indeksi ile ilişkisi incelendiğinde ise, yalnız 6.5 derece için anlamlı ilişki bulunmuştur. Katılımcıların kol çevirme hızları vücut ağırlığı dolayısıyla kolların ağırlığı (r squared= 0,83), boy uzunluğu (r squared= 0,50), vücut kitle indeksi (r squared= 0,76) ve kolların eylemsizlik momenti (r squared= 0,74) arttıkça düşmektedir.

Tartışma

Bu çalışmada bireylerin denge kaybının şiddetine bağlı olarak toparlanma amacıyla kollarını ne kadar etkili kullandıkları ve farklı açılarda ne tür kinetik ve kinematik değişiklikler olduğu araştırılmıştır. Katılımcıların 5.5, 6.5 ve 7.5 derecelerde duvara sabitlenmiş bir halat yardımıyla öne doğru eğilmeleri sağlanmış ve aniden serbest bırakılmışlardır. Katılımcılardan dengelerini yeniden sağlamak için adım almadan önce vücut uzuvlarını kullanarak toparlanmaya çalışmaları istenmiştir. Katılımcıların tamamı kollarını öne doğru çevirerek toparlanmaya çalışmışlar fakat farklı çevirme stratejileri kullanmışlardır. Katılımcıarın hiçbirisi 7.5 dereceden toparlanmayı başaramamıştır.

Toparlanma sayıları incelendiğinde hemen hemen tüm katılımcıların 5.5 dereceden toparlanmayı başardıkları söylenebilir. Toplam 50 testten 47sinde toparlanma gerçekleşmiştir. 6.5 derece incelendiğinde ise 50 testten 30'unda toparlanabildikleri görülmüştür. Literatür incelendiğinde genç bireylerin maksimum toparlanabilme açıları 7.2 derece olarak rapor edilmiştir (D. C. Mackey & Robinovitch, 2006). Bu da bizim katılımcılarızın 7.5 dereceden toparlanamamalarını açıklamaktadır. Ancak Macket ve Robinovic yaptıkları çalışmada denekerden yalnızca kaslarını kullanmalarını istemişlerdir. Bu çalışmada EMG ölçümü alınmadığı için katılımcıların kaslarını ne kadar etkili kullanıkları yönünde herhangi birşey söylemek imkansızdır. Yalnız kolların kaslarını etkili kullandıklarını varsaysak bile 0.3 derecelik açığı kapatmaya yetecek kadar etkili olmadığını söyleyebiliriz. Kolların kullanımı 7.5 dereceden toparlanabilmek için yeterli gelmemektedir.

Kollarda oluşan açısal momentum iki farklı şekilde hesaplanmıştır. Yukarıda da açıklandığı gibi birincisi omuz eklemi baz alınarak hesaplamada kolların diğer vücut uzuvları dikkate alınmadan kolların kendi başlarına yaptıkları etki incelenmiştir. Burada da iki farklı etkiden söz edebiliriz. Birincisi kolların vücudun önünde ve aşağıda doğru çevirilmesi ile oluşan negatif (koordinat düzlemine göre) açısal momentum. Bu momentum yere ayak parmak uçlarının daha kuvvetli bir şekilde baskı uygulamasını sağlarken aynı zamanda kolların vücudun önünde vücut ağırlık merkezinin öne doğru kaymasına ve doğal olarak düşme olasılığını arttırmasını engellemek amacıyla vücudun arkasına daha hızlı alınmasını sağladığı düşünülmektedir.

Katılımcılar serbest bırakıldıklarında kolların ilk hareketi yanlara ve geriye doğru açılmasıdır. Bu refleksif hareker ağrılık merkezinin geriye doğru alınması amacıyla yapılıyor olabilir. Bu uyaranın verilmesinden kısa bir süre içerisinde oluştuğu için planlamış bir hareket olma olasılığı düşüktür. Kolların ilk harekete başladığı zaman 0.28-0.30 sn olarak bulunmuştur. Bu da kasları bilinçli bir şekilde aktive edebilmek için yeterli bir zaman değildir. Kasların bilinçli bir şekilde beynin motor korteks kısmı tarafından yönetilmesi için yaklaşık 0.400-0.600 sn gibi bir süre gerekmektedir.

Kolların refleksif olarak geriye doğru alınması ve vücut ağırlık merkezinin denge alanı içerisinde tutulmasının ardından bilinçli olarak yapılan harekete geçmek için zaman kazanılmış olur. Bu aşamada da katılımcıların tümünün kollarını öne doğru çevirdikleri görüşmüştür. Kolların vücudun önünde aşağı doğru hareketi sırasında oluşan açısal momentumlar 5.5 ve 6.5 dereceler için karşılaştırıldığında 6.5 derecede oluşan açısal momentumun istatistiksel olarak yüksek olduğu görülmektedir. Pozitif açısal momentumun oluştuğu kolların vücudun gerisinde yukarı doğru çevirilmesi ile oluşan açısal momentumlar arasında anlamlı farklılık bulunamamıştır.

İkinci açısal momentum kolların ayak parmak uçlarının orta noktasına göre hesaplanan açısal momentumdur. Bu da kolların diğer vücut uzuvları dikkate alınarak etkilerinin anlaşılması amacıyla hesaplanmıştır. Burada da hem pozitif hem de negatif açısal momentumlar için 5.5 ve 6.5 dereceler arasında anlamlı farklılık bulunmuştur. Bu sonuç da katılımcıların denge kaybının ardından toparlanabilmek için kollarını daha etkili kullanmaya çalıştıklarını göstermektedir.

Yukarıda belirtilen farklılıkların nereden kaynaklandığını tespit etmek amacıyla kolların çevirilme hızları hesaplanmıştır. Buradaki önemli soru acaba katılımcılar toparlanabilemek için kolarını 6.5 derecede daha hızlı çevirerek mi açısal momentumlarını arttırdığıdır. Yapılan istatistiksel analizler göstermiştir ki katılımcılar kollarını 6.5 derecede çok daha hızlı çevirmiştlerdir. Bu da bize denge kaybı sırasında toparlanabilmek için kollar rastgele veya çevirilebildiği kadar hızlı değil, toparlanabilmek için yeteri kadar hızlı çevirildiğini göstermektedir.

Kolların açısal hızlarının farklı değişkenler ile ilişkisi incelendiğinde ise şu sonuçlar bulunmuştur. Vücut ağırlığı arttıkça katılımcıların kol çevirme hızları düşmektedir. Vücut ağırlığının artması kolların ağırlıklarının da artması anlamına gelmektedir. Bireyin kol ağırlığının fazla olması aynı miktarda açısal momentum oluşturmak için kollarını, kolları zayıf olan bireylere göre daha yavaş çevirmesi yeterli olacaktır. Daha ağır katılımcılar kollarını daha yavaş çevirmişlerdir. Aynı sonuçlar vücut kitle indeksi, boy uzunluğu ve kolların eylemsizlik momenti için de bulunmuştur. Temel olarak bu değişkenlerin hepsi aynı sonuçu göstermektedir.

Bireylerin kollarını denge kaybının şiddetine bağlı olarak arttırmaları, toparlanabilmek için merkezi sinir sisteminde birtakım hesaplamaların yapıldığını göstermektedir.

Bu sonuçlar aynı zamanda sportif aktiviteler sırasında dengenin korunması veya vücut uzuvlarının açısal momentumun korunmasında ne kadar önemli olduğunu göstermektedir.

Sonuç ve Öneriler

Bu çalışmada, denge kaybının şiddeti arttıkça toparlanabilmek için kolların daha etkin kullanıldığı bulunmuştur. Kollar refleksif olarak çevirilebildiği kadar hızlı şekilde değil, bireyin kollarının ağırlığı ve kolların eylemsizlik momentine bağlı olarak çevirildiği bulunmuştur.

Bu çalışmanın en önemli hipotezi, denge kaybının şiddeti arttıkça kolların açısal momentumlarının, açısal hızlarının, yere uygulanan dikey kuvvetin, toparlanma sürelerinin ve omuz torklarının farklılık göstereceği yönündeydi. Katılımcıların 5.5, 6.5 ve 7.5 derecelerde öne doğru eğik bir şekilde durmaları sağlandı. Daha sorna serbest bırakılarak adım almadan önce vücut uzuvlarını kullanarak toparlanmaları istendi. Yapılan analizler sonucunda farklı şiddetteki denge kayıpları arasında kolların açısal momentumları ve açısal hızları arasında anlamlı farklılıklar bulundu. Katılımcılar denge kaybının şiddeti arttıkça kollarını daha hızlı çevirdiler. Aynı zamanda kolları çevirme hızları katılımcıların vücut ağırlıklarına, boylarına, kollarının eylemsizlik momentlerine ve vücut kitle indekslerine bağlı olarak değişiklik gösterdi. Bireylerin, kollarının ağırlıkları, boyları, vücut kitle indeksleri ve kolların eylemsizlik momentleri arttıkça kollarını daha yavaş çevirdikleri bulundu. Bu da aynı miktarda açısal momentum etkisini yaratabilmek için merkezi sinir sisteminin birtakım hesaplamalar yaparak vücut hareketlerini kontrol ettiğini göstermektedir.

Bu çalışma sonucunda bulunanlar, sportif aktiviteler sırasında, özellikle yüksek denge becerisi gerektiren cimnastik, kayak ve artistik sporlar ve ayrıca teknik becerilerin yüksek olması gereken sporlar sırasında vücut uzuvlarının etkili kullanımının, tekniği doğru ve etkili uygulayabilmek için ne kadar önemli olduğunu göstermektedir. Ancak bu sonucu desteklemek için karmaşık hareketler içeren spor branşlarında ve tekniklerde bu tür araştırmaların yapılması son derece önem taşımaktadır.

I. TEZ FOTOKOPİSİ İZİN FORMU

ENSTİTÜ

Fen Bilimleri Enstitüsü	
Sosyal Bilimler Enstitüsü	X
Uygulamalı Matematik Enstitüsü	
Enformatik Enstitüsü	
Deniz Bilimleri Enstitüsü	

YAZARIN

Soyadı : AK
Adı : EMRE
Bölümü : Beden Eğitimi ve Spor Bölümü

TEZİN ADI : Investigation of the Use of the Arms in Recovering from Postural Perturbations

TEZİN TÜRÜ : Yüksek Lisans [ ] Doktora [X]

1. Tezimin tamamından kaynak gösterilmek şartıyla fotokopi alınabilir. [X]
2. Tezimin içindekiler sayfası, özet, indeks sayfalarından ve/veya bir bölümünden kaynak gösterilmek şartıyla fotokopi alınabilir. [X]
3. Tezimden bir (1) yıl süreyle fotokopi alınamaz. [ ]

TEZİN KÜTÜPHANEYE TESLİM TARİHİ: