Aging Effect on Successful Reactive-Recovery from Unexpected Slips: a 3D Lower Extremity Joint Moment Analysis

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ABSTRACT

The objective of the proposed study was to perform three-dimensional (3D) inverse dynamics analysis to determine lower extremity (ankle, knee and hip) joint moments on previously collected slip perturbation experimental data. In addition, the aging effect on joint moment generation in both normal walking and reactive-recovery conditions was examined.

Dataset collected during previous slip and fall experiments, which were conducted in a typical gait analysis setting, were analyzed in current study. All the participants were subjected to the screening criteria, which defined the successful reactive-recovery (i.e. non-fall trials) based on slip distance, sliding heel velocity, whole body COM velocity, and motion pictures. Nine young and nine old healthy participants, who were identified possessing representative trials, were involved as participants in current study.

A local coordinate system was constructed on each joint and each segment of the lower extremity based on available landmarks using the Gram-Schmidt orthogonalization algorithm. 3D inverse dynamics was implemented to obtained lower extremity joint moments. Magnitude and timing of obtained joint moment patterns during stance phase were subjected to one and two-way analysis of covariance (ANCOVA) with walking velocity as covariate. The aging effect and gait condition effect were evaluated.

Increases in peak joint moment, peak joint power, and joint moment generation ratio were detected in successful reactive-recovery. Distinct age-related joint moment generation strategy was observed through findings of peak joint moment ratio and joint

moment generation rate. The elderly, who were able to reactive recover, were found to be as rapid as their younger counterparts in terms of initiating and developing reactive joint moment.

It was concluded that the ankle joint was critical in balance recovery while the hip joint assumed the major responsibility of balance maintenance of upper body during successful reactive-recovery. Increased demand on muscle strength during balance recovery lead to the distinct joint moment generation strategy adopted by the elderly, and confirmed the necessity of lower extremity strength training to allow higher joint moment generation. In addition, implementation of 3D joint moment analysis was justified in current study and was recommended in future slip and fall research.

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1. INTRODUCTION

1.1. Rationale

Falls resulting from unsuccessful recovery responses to destabilizing events (slips) are of particular concern in the elderly individuals. Reactive recovery responses are critical in maintaining dynamic stability following an unexpected slip event and thus it is important to understand the nature of reactive-recovery.

The human body will undertake a series of significant age-related changes in the neurological, physiological, and the musculoskeletal system. Of particular changes associated with aging, joint moment received much attention. Joint moment patterns give valuable insight into the net effect of all agonist and antagonist muscle activity, and thus represent an integration of the neural control acting at each joint (Winter, 1980). Studies suggest that in normal walking, elderly people cannot generate ankle joint moments as rapidly as the younger group (Thelen, Schultz, Alexander, & AshtonMiller, 1996). Explosive strength generation and the ability to attenuate fast, large-scale lower extremity motions are proposed to be critical in determining whether or not a person can respond appropriately to balance perturbation (Do, Breniere, & Brenguier, 1982). Therefore, diminished rapid joint moment development capacities of older adults may predetermine their available balance recovery strategies in the event of a slip, and may increase likelihood of fall accidents.

In terms of slip-induced fall injuries, several studies have contributed to reveal the aging effect in human reactions to slippery surfaces (Lockhart, Woldstad, Smith, & Ramsey, 2002; Lockhart, Woldstad, & Smith, 2003; Lockhart, Woldstad, & Smith, 2002). Although much has been learned about the kinematic and physiological changes associated with aging in slips and falls, few studies have investigated aging effects on lower extremity joint moments. Considering the necessity of thorough understanding of slips and falls associated with the elder population, it is necessary to study the aging effect on joint moments during reactive-recovery events.

Joint moment analyses are usually accomplished using an inverse dynamics approach, which can be performed in two-dimensional (2D) or three-dimensional (3D) space. Both 2D and 3D joint moment analyses in normal gait have been well documented in the literature. Despite the more complex experimental setting and more intensive computation required by 3D analysis compared to 2D analysis of joint moments, the application of 3D joint moment analysis is warranted as it reveals more information to help understand moment generation mechanism, especially for the hip joint.

Two dimensional inverse dynamics approach has been used in several studies to determine joint moments in an attempt to reveal the causes of slip and fall accidents. Ferber et al. (2002) simulated natural slip events using unexpected forward transitional perturbations. Distinct 2D lower extremity joint moment patterns were observed. Cham and Redfern (2001) investigated lower extremity corrective reaction when walking across the unexpected slippery surface. Their examination of 2D lower extremity joint moment revealed different response patterns compared to experiments simulated using forward perturbation.

In summary, 3D joint moment analyses is warranted and expected to provide further insight of human reactions to unexpected slippery surface. Therefore, to better understand age-related reactions to slip events through joint moment analysis, 3D inverse dynamic approach was applied in this proposed study.

1.2. Research Objective

The overall objective of current study was to investigate aging effects on lower extremity (ankle, knee and hip) joint moments determined during successful reactive-recovery from unexpected slips.

1.2.1. Hypotheses related to timing variables

When individuals are exposed to slippery surfaces during level walking, being able to respond in a relatively short time period is critical to successfully recovery from slips. Previous literatures suggest that the elderly cannot generate joint moments as rapidly as their younger counterparts (Thelen et al., 1996). It is also proposed that the likelihood of recovery from a trip may be increased if the response time is decreased (van den Bogert, Pavol, & Grabiner, 2002). The lack of such ability in the elderly may contribute to the cause of age-related fall injuries. Thus, investigation of timing differences may contribute to the knowledge of the safe time which is not too late for an individual to actively recover from slips. It is intuitive that response time will be longer in falls than in successful reactive-recovery. Even solely in the reactive-recovery condition where the elderly also regained their balance successfully as their younger counterparts, the differences in response time influenced by aging may be still present, which may indicate the hidden risk for the elderly. It was then hypothesized that the elderly would need more time to initiate and to generate sufficient joint moments, which was hypothesized to be closely correlated with peak joint moment after slip initiation, to successfully recover from unexpected slips than their younger counterparts.

In current study, two types of timing variables were determined and analyzed to test this hypothesis. The first was the reactive joint moment activation time (JMA Time). This variable was used to characterize the time needed to initiate reactive joint moment responding to slip events. The activation threshold of reactive joint moment was defined as the average joint moment plus 2 SD (standard deviation) during the period from slip start to slip end. JMA Time was then defined as the duration from heel contact to activation of reactive joint moment. Specifically, the elderly was hypothesized to have longer JMA Time than their younger counterparts.

The second timing variable was the joint moment activation to peak time (JMAP Time). This variable was used to characterize the time needed to generate peak joint moment required for the successful reactive-recovery. JMAP Time was defined as the duration from the joint moment activation to the peak joint moment. Specifically, the elderly was hypothesized to have longer JMAP Time than their younger counterparts.

1.2.2. Hypotheses associated with magnitude variables

Another factor that may affect the outcome of human reactions to slips is the joint moment magnitude. Successful reactive-recovery from unexpected slips requires individuals to generate sufficiently large joint moments to regain dynamic balance. Thus, it was hypothesized that individuals would generate more joint moment during reactive-recovery than normal walking in both age groups.

In addition, the elderly are known to have decreased joint moment generation capacity due to the age-associated musculoskeletal degradation. During reactive-recovery, the joint moments that the elderly generate may be far beyond their lower extremity strength measurement. It was hypothesized that the ratio (peak joint moment to the corresponding maximum muscle strength) generated by the elderly would be larger than the ratio produced from their younger counterparts during successful reactive-recovery.

In current study, the peak joint moment magnitude (JMP Magnitude) was used to characterize the moment generation difference between normal walking and reactive-recovery from slips. The JMP Magnitude was defined as the peak joint moment from slip start to toe off. It was specifically hypothesized that the JMP Magnitude would be higher in reactive-recovery than in normal walking for both age groups.

The joint moment generation ratio (JMG Ratio) was used to characterize the ability of moment generation during reactive-recovery period. JMG Ratio at ankle and knee joints was defined as the peak sagittal joint moment magnitude in one joint to the maximum isokinetic muscle torque measurement at that joint. It was specifically hypothesized that the JMG Ratio would be larger for the elderly group than the younger group in the reactive-recovery condition.

The peak joint power magnitude (JMP Power) will help to characterize the role of muscle groups acting about a joint. Joint Power at the ankle, knee and hip joints is determined as the product of joint angular velocity and net joint moment (Riley, DellaCroce, & Casey Kerrigan, 2001). Mechanical power is the single variable that summarizes the function of muscles as they shorten and lengthen under tension (Winter, Eng, & Ishac, 1995). Younger participants were expected to utilize their lower extremity energy generation or absorption more than the elderly in the condition where fast balance maintaining was required. Thus, it was hypothesized that the JMP Power would be lower for the elderly group than the younger group in the reactive-recovery condition.

1.2.3. Hypotheses associated with ratio

In the current study, 3D inverse dynamics was applied to determine joint moments in three reference planes (frontal, sagittal and transverse) in an attempt to provide complete information regarding the aging effect on joint moments during reactive-recovery period. Studies have shown that in normal walking, the majority of joint moment is generated in sagittal plane for forward progression of the whole body Center-of-Mass (COM), while lower joint moments are generated in frontal and transverse planes for balance adjustment. However, in slip and fall situations where dynamic balance is perturbed and regained rapidly, balance control functionality of the lower extremity are believed to be actively involved. Therefore, it was hypothesized that significantly large joint moments will exist in all three reference planes during successful reactive-recovery.

The peak joint moment ratio (JMP Ratio) was used to characterize the distribution of peak joint moments on each of the three reference planes for specific joints and the relative importance of each reference plane in terms of successful reactive-recovery. Due to the fact that individuals may utilize higher joint moments in all three reference planes in the process of maintaining dynamic stability, ratio variables were expected to reveal the information of relative contribution of each reference plane. JMP Ratio for each plane was defined as the percentage of the peak joint moment in that plane relative to the summation of the peak joint moments in all three reference planes for specific joint. As relative joint moment contributions by frontal and transverse plane muscle groups increase, sagittal plane joint moment relative contribution will decrease. Therefore, it was hypothesized that sagittal plane JMP Ratio for each joint would be smaller in reactive-recovery than in normal walking. By confirming this hypothesis, current study would increase our knowledge of the role of frontal and transverse plane muscle groups crossing each joint in reactive-recovery.

The joint moment generation rate (JMG Rate) was used to characterize the speed of peak joint moment generation. JMG Rate for each plane at each joint was defined as the slope of peak joint moment in that plane relative to the corresponding JMAP Time. Since the elderly may produce smaller peak joint moment magnitude than their younger counterpart, JMG Rate will help to measure how fast the joint moments develop once participants encounter external perturbations such as slippery surface. It was hypothesized that JMG Rate in each reference plane at each joint would be higher for the younger group than for the elderly group during reactive recovery process due to aging effect on musculoskeletal system.

1.2.4. Summary

In summary, the following hypotheses were tested in current study:

a) The elderly would have longer JMA Time (as of ankle, knee and hip) than the younger individuals during successful reactive-recovery.

b) The elderly would have longer JMAP Time (as of ankle, knee and hip) than the younger individuals during successful reactive-recovery.

c) JMP Magnitude (as of ankle, knee and hip) would be larger during successful reactive-recovery than during normal walking for both young and old.

d) The elderly would have larger JMG Ratio (as of ankle and knee) than the younger individuals during successful reactive-recovery.

e) The sagittal plane JMP Ratio (as of ankle, knee and hip) for each lower extremity joint would be smaller during reactive-recovery than during normal walking for both age groups.

f) The elderly would have lower JMG Rate (as of ankle, knee and hip) than the younger individuals during successful reactive-recovery.

g) The elderly would have lower JMP Power (as of ankle, knee and hip) than the younger individuals during successful reactive-recovery.

1.3. Need for this Study

Improved understanding of age-related human reactions in reactive-recovery through joint moment analysis, which reveals the underlying mechanical cause of the observed kinematics, would enhance our ability to identify the successful reactive-recovery mechanisms involved in slip and fall accidents.

C h a p t e r 2

2. LITERATURE REVIEW

Slip-induced fall accidents are a major cause of serious injuries and deaths. It is estimated that falls cause 17% of all occupationally-related injuries and 18% of injuries in the public section in the USA (Leamon & Murphy, 1995). The annual direct cost from occupational injuries due to slips, trips and falls in the USA has been estimated to exceed $6 billion (Sorock, Lombardi, Courtney, Cotnam, & Mittleman, 2001). According to the Bureau of Labor Statistics (1999, 2000, 2001), floors and walkway or ground surfaces were identified as the major sources of slip and fall accidents, causing over 86% of all fall-related injuries. The elderly population is especially susceptible to slip and fall injuries. Statistics show that approximately one-third of adults over 70 years of age fall in a given year, with one-fourth of those falls resulting in fall-related injuries (Centers for Disease Control and Prevention, 2000). Understanding the internal factors that cause slip and fall accidents and their complex interaction with environmental factors would be extremely helpful in searching for successful intervention solutions.

This literature review will consist of three parts. The first part deals with current biomechanical analysis of slips and falls. 3D joint moment analysis is reviewed in the second part, while the aging effect in gait study is described in the third part.

2.1. Current biomechanical analysis of slip and falls

Biomechanical analysis usually involves kinematic and kinetic analyses. Kinematics, by definition, is the description of human movement independent of the forces that cause the movement (Winter, 1979). They include linear and angular displacements, velocities, and accelerations. Extensive kinematic studies have been conducted to describe and categorize slip and fall events. Some parameters extend from traditional gait parameters. For example, Bunterngchit et al (2000) studied stride length between walking trials on normal and slippery surfaces. Brady et al (2000) studied the possibility of recovery from an induced slip by investigating foot heel-strike angles. Cham et al (2002b) investigated heel dynamics including heel velocity, heel acceleration and foot angle at the time when heel contacts the slippery surface. Lockhart et al (2003) included center of mass (COM) velocity as one of the important gait parameters in an attempt to reveal gait changes during slips and falls. Other interesting kinematic parameters are created to facilitate the description of slip and fall accidents. For example, slip distance I and slip distance II were specially used to characterize the severity of slips and falls (Lockhart et al., 2002). Slip distance I was obtained using the horizontal heel coordinates between slip-start (the point where non-rearward heel acceleration occurs after heel contact) and mid-slip point (the point where peak horizontal heel acceleration occurs after slip-start). Slip distance II was determined from the horizontal heel coordinates between mid-slip point and slip-end point (the point where the first maximum of the horizontal heel velocity occur after slip-start).

Beyond description and categorization of slips and falls, kinetic analyses are extremely powerful for a deeper understanding of the cause of slip and fall accidents. Kinetic analysis is defined as the study of the forces that cause the movement (Winter, 1979). It usually involves the studies on ground reaction force (GRF), joint moments, joint work, joint power as well as muscle force.

In slip and fall studies, GRF related variables have been widely explored. It was stated that most of slip-induced falls occurred when the frictional force (F_μ) opposing the direction of foot movement is less than the shear force (F_h) of the foot immediately after the heel contacts the floor (Perkins & Wilson, 1983). Lockhart (2000) used the required coefficient of friction (RCOF) to study the potential for slip-induced falls between age groups. The RCOF is the ratio of horizontal ground reaction force to vertical force and represents the minimum requirement on the available coefficient of friction between the shoe and floor interface to prevent slipping (Perkins, 1978). It has been commonly recommended by standards organizations and by individual authors that static COF should be 0.5 on level walking surfaces (Lin, Chiou, & Cohen, 1995).

Based on the description of kinematic parameters, human reactions to slippery surfaces can be categorized into no-slip, reactive-recovery, and fall trials (Cham & Redfern, 2001; Strandberg & Lanshammar, 1981; Perkins, 1978; Hanson, Redfern, & Mazumdar, 1999). There is no general agreement on the definition of these three categories. Several researchers solely rely on the distance (Perkins, 1978; Leamon & Li, 1990; Gronqvist, 1999). For example, Perkins (1978) define trials with slip distances less than 1 cm as no-slip, trials with slip distances between 1 to 10 cm as reactive-recovery, and the other trials as falls. Some studies combined slip distance along with peak forward velocity to categorize human reactions more precisely (Strandberg, 1983; Cham & Redfern, 2002b). Furthermore, Brady et al (2000) utilized the force measured from the safety harness load cell to assist classification of each slip outcome as a recovery of a fall.

While relevant kinematic studies and understanding of RCOF are important in terms of description and categorization, kinetic studies are necessary to understand the causes associated with slip-induced fall accidents. Due to the complexity of joint kinetics computation and the difficulty to reproduce the unexpected nature of real-life slipping accidents (Cham & Redfern, 2002a), few studies investigating the lower extremity joint kinetics during slip-induced fall accidents are available in the literature. Ferber (2002) simulated slip event using unexpected forward perturbations to assess the corresponding joint moments and joint powers. They concluded that the hip muscles were the most important muscles associated with maintaining balance while experiencing such events. Distinct lower extremity joint moment pattern was also observed. Cham and Redfern (Cham et al., 2001) examined lower extremity joint moments for experimental evidence of corrective strategies during a slipping event. It was found that the corrective strategies included increased knee flexion moment and hip extensor moment. The ankle joint, on the other hand, was found to act as a passive joint during fall or reactive-recovery trials.

Therefore, several slip and fall studies have been performed from the perspective of joint kinetics. Further investigations on joint kinetics on slip and fall accidents, especially a more comprehensive analysis of the joint kinetics required for successful reactive-recovery strategies adopted in response to unexpected slips would be very helpful to expand our knowledge of slip and fall accidents.

2.2. 3D joint moment analysis in gait study

One of the most valuable kinetic variables associated with the assessment of human movement is the time history of the lower extremity joint moments (Winter, 1980). Net joint moment is defined as the product of net muscle force times the moment arm measured from joint center to segment center (Winter, 1979). Joint moment patterns give valuable insight into the net effect of all agonist and antagonist muscle activity, and thus represent an integration of neural control acting on each joint (Winter, 1980). In gait studies, joint moment analyses mainly focused on lower extremity joints (i.e. ankle, knee, and hip) (Allard, Lachance, Aissaoui, & Duhaime, 1996; Apkarian, Naumann, & Cairns, 1989; Alkjaer, Simonsen, & Dyhre-Poulsen, 2001; Ounpuu, Davis, & DeLuca, 1996).

Inverse dynamics is the most commonly used algorithm to determine lower extremity joint moments. Estimation of joint moment patterns via inverse dynamics has been applied widely for various types of activities. Winter (1983) analyzed the joint moment patterns from slow jogging trials. Arampatzis et al (1999) investigated the influence of speed on maximum joint moments in human running. Lundin et al (1995) studied the bilateral lower extremity joint moment sysmmetry during sit-to-stand motion. Beyond testing the healthy population, researchers also utilized inverse dynamics to reveal the effect of pathological gait on lower extremity joint moments (Ounpuu, Gage, & Davis, 1991; Ounpuu, 1995; Ounpuu et al., 1996; Lai, Kuo, & Andriacchi, 1988).

According to the dimensions studied in 3D space, joint moment analysis can be categorized into 2D and 3D analysis. 2D joint moment analysis is a simplified approach, which requires only one camera to record the movement, and few markers placed on anatomical landmarks to define the joint center position in 2D space (Alkjaer et al., 2001). 2D joint moment analysis is based on the assumption that body segment motion occurs within a vertical plane that is parallel to the direction of forward progression, referred to as the 'sagittal plane', in most types of activities (Winter, 1980). Motion in the transverse plane, the frontal plane, is small in magnitude compared to the sagittal plane during walking (Novacheck, 1998).

3D joint moment analysis, on the contrary, is a complete analysis focusing on all the three reference planes. The necessity for doing 3D analysis comes from the fact that sagittal kinetic analysis provides only part of the information, particularly at the hip joint where hip abductors are critical for the balance control of the trunk in the frontal plane (Mackinnon & Winter, 1993). Furthermore, the accuracy of motion description especially of the hip joint, deteriorates as the number of degrees of freedom considered is reduced due to the simplifying 2D assumptions (Cappozzo & Gazzani, 1990). It was found that even almost planar movements (such as walking and running) were associated with significant 3D (especially in the frontal plane) intersegment moments (Glitsch & Baumann, 1997). Therefore, when the internal loads, such as joint moments, of the body structures are the focus of interest, each step (calculating both kinematic and kinetic quantities) of the inverse dynamics has to consider three dimensions.

3D joint moment analysis has been well documented in the literature. Kadaba et al (Kadaba et al., 1989) and Apkarian et al (Apkarian et al., 1989) described the 3D joint moments generated during normal walking of healthy participants. Eng et al (1995) reported lower extremity joint moment patterns as well as muscle power. Manal et al (2002) investigated the effect of soft tissue movement of the shank on the 3D knee moment profiles. Allard et al (1996) performed a bilateral symmetry study on 3D joint moment, power, and work produced by young male participants. Glitsch et al (1997) calculated 3D joint moments as the input of their optimization algorithm to estimate muscle force.

Meanwhile, analyses of 3D joint moment patterns have also been widely applied to assist pathological gait studies. Houck and Yack (2003) compared knee angles and moments of healthy participants and participants with anterior cruciate ligament deficiency (ACLD) during stepping and crossover cutting activities. It was found that primarily knee frontal and transverse plane moments distinguished the stepping and crossover cut activities. Hurwitz et al (1998) examined relationship of the loads at the hip joint during gait and the bone mineral density of the proximal femur in patients with end-stage hip osteoarthritis, with the assistance of 3D hip joint moment patterns. In their study, hip frontal and transverse plane moments were found significantly correlated with bone mineral density.

In addition to its valuable direct application, joint moment is also the first stage of calculating several other joint kinetic variables (joint power, joint work, and joint stiffness) theoretically. Joint power, which represents the efficiency of energy flow at a joint due to the contribution of muscles acting across about a joint (Riley et al., 2001), is calculated from the scalar product of the joint angular velocity and joint moment (Robertson & Winter, 1980). Joint work is calculated as the time integral of the power curves (Eng & Winter, 1995). Dynamic joint stiffness represents as a measure of resistance to change in the joint angle (Hunter & Kearney, 1982). Joint stiffness can be determined as the gradient of the joint moment versus joint angle graph (Lark, Buckley, Bennett, Jones, & Sargeant, 2003).

In summary, complete determination of joint moments in three-dimensional (3D) space using inverse dynamics approach has been applied widely and proven to be a valuable tool in normal and pathological gait analysis. Successfully 3D joint moment estimation will also enable further investigation of additional kinetic parameters.

2.3. Aging effect in gait study

The human body will undertake significant neurological, physiological, and musculoskeletal changes as age increases. Age-related changes in the nervous system include increased brain mass loss rate, increased reaction times, decreased neurotransmitter production and a decreased acuity of the auditory, vestibular, visual and somatosensory systems (Peterka, Black, & Schoenhoff, 1990; Hayes & Jerger, 1984; Rogers & Bloom, 1985). Physiological changes associated with aging include the general decreased muscle strength and loss of passive range of joint motion (Winter, 1991; Trueblood & Rubenstein, 1991; Bendall, Bassey, & Pearson, 1989). In general, isometric and isokinetic muscle strengths peak in the mid-twenties and then decreases slowly until after 50 years of age when there is an accelerated decline (Frontera et al., 2000; Frontera, Hughes, Lutz, & Evans, 1991).

The aging effects on gait parameters are evident in both kinematic and kinetic aspects. Kinematic changes associated with aging can be summarized as the decreased stride length, cadence and walking velocity. Winter (1991) and Oberg et al. (1994) reported decreased ankle dynamic range of motion and increased hip dynamic range of motion in the elderly compared to their young counterparts. Meanwhile, higher heel contact velocity, lower hip acceleration and greater head acceleration were found in the elderly (Winter, 1991).

A limited amount of literature focuses on kinetic comparisons between the young and the elderly in normal gait. Winter (1991) showed that antero-posterior peak GRF during push-off is less (1.93 N/kg) in the elderly than in the young (2.19 N/kg), which represented a less push-off. Several joint moment profiles in both young and elderly were reported (Winter, 1991; Judge, Davis, & Ounpuu, 1996; Thelen et al., 1996). Winter (1991) found lower peak ankle plantarflexor moment in the elderly than in the young. Judge et al (1996) reported a much smaller peak ankle plantarflexor power in the elderly. A study by Thelen et al (1996) showed that the elderly cannot develop ankle joint moment as rapidly as the young, which was thought to be essential for quick recovery following balance perturbations.

In summary, previous research in gait studies have drawn attention to both kinematic and kinetic changes associated with aging. With the increasing life expectancy and active lifestyle of the elderly, determination of any aging-related characteristics, especially in kinetic domain, will be warranted in order to identify diagnostic measures and develop appropriate intervention programs.

2.4. Summary

Slip-induced fall accidents are a major cause of serious injuries and deaths for the elderly. Complete 3D lower extremity joint moment analysis of reactive-recovery responses will not only increase our knowledge of strength-related factors associated with dynamic balance maintaining in slip and fall accidents, but also aid in developing strength focused intervention programs. Meanwhile, with consideration of aging effect, this proposed study will also have particular implications regarding the age-related balance recovery from unexpected disturbance as in slip and fall accidents.

Chapter 3

3. METHODS

Previously collected (detailed procedure will be described) data in slip and fall experiments conducted in the Locomotion Research Lab at Virginia Tech was utilized. Three-dimensional lower extremity biomechanical models were constructed according to available marker configurations (specific setting for slip and fall experiment). Segment fixed (local) coordinate systems were constructed according to Gram-Schmidt algorithm (Bradley, 1975). Segment linear and angular kinematic quantities were calculated based on available external landmark position data. The participant's anthropometric measures and published anthropometric distributions were used as anthropometric input. Standard three-dimensional inverse dynamics was performed to calculate joint moments in all three reference planes (frontal, sagittal and transverse).

3.1. Participants

3.1.1. Entire participants group

Thirty-two young (19-35 years old) individuals (17 males and 15 females) and thirty elderly (68-86 years old) individuals (15 males and 15 females) participated in the slip and fall experiment (conducted in 2002 and 2003). Informed Consent was approved by the Institutional Review Board of Virginia Tech and was obtained from all the participants. Medical history forms filled by the participants indicated that they were physically healthy.

3.1.2. Participants screening criteria

As stated in the literature review section, human reactive responses when walking across slippery surfaces vary from successful recovery and fall. Thus, target data trials analyzed in current study were identified from the entire datasets before further data processing.

Successful reactive-recovery was identified as trials with participants successfully walking over slippery floor surface without falling. Slip distance, sliding heel velocity, the whole body COM velocity, and motion pictures were considered to identify the fall events. To be considered as a fall, the slip distance must exceed 10 cm, and peak sliding heel velocity must exceed the whole body COM velocity while slipping (Lockhart et al., 2002). Also, videos for each participant were analyzed to see if falls had occurred (the participants lost balance and the fall was arrested by the harness). All other trials on the slippery floor surfaces were identified as reactive-recovery trials. Furthermore, the participants who successfully recovered with two feet (i.e., with swing foot) on one force-plate were not used in current study.

3.1.3. Sample size estimates

Estimation of required sample sizes for the experiment proceeded from estimates of intersubject variability in peak ankle plantar-flexor muscle moment obtained from previous study (Khuvasanont, 2002). Power of tests was determined by focusing on sample sizes large enough to detect differences between younger and older participants with high probability. The formula for determining the power of the test (Neter, Kutner, Nachtsheim, & Wasserman, 1996) is given by:

$$Power = P\left\{\left|t^*\right| > t\left(1-\alpha/2; n-2\middle|\delta\right)\right\} \tag{3.1}$$

where δ is the noncentrality measure, which is a measure of the distance between the means of *A* and *B* (peak ankle plantar-flexor muscle moment of younger and older participants):

$$\delta = \frac{|A-B|}{\sigma\sqrt{2/n}} \tag{3.2}$$

where σ is the standard deviation of the distribution of peak ankle plantar-flexor muscle moment and *n* is the number of participants in each age group.

The difference between *A* and *B* (i.e. the minimum difference which is important to detect with high probability) was assumed to be 0.08 Nm/kg (Khuvasanont, 2002), and the standard deviation of peak ankle plantar-flexor muscle moment was approximately 0.06

Nm/kg (Khuvasanont, 2002). Specifying that α =0.05, 9 participants in each of the age groups should be sufficient to detect the specified differences in lower extremity joint moments with risks of Type I error of 0.05 and Type II error of <0.25 (Power >0.75).

3.1.4. Participants in current study

Nine young participants (19-35 years old) and nine older participants (67-79 years old) were identified as possessing reactive-recovery trials during data processing. All the participants claimed no history of significant musculoskeletal and neurological disease and injury. The mean and standard deviation of age, weight and height of all the participants were presented in Table 3.1-1. In addition, all the participants reported their right feet as dominant feet.

Table 3.1-1 Participant Information (by age)

	Young (20-35 years old)	Old (68-78 years old)
	Mean (S.D.)	Mean (S.D.)
Age (yr)	23.56 (4.77)	73.56 (4.42)
Weight (kg)	65.48 (8.41)	74.60 (12.57)
Height (cm)	167.27 (10.34)	166.2 (6.98)

3.2. Data collection

3.2.1. Apparatus

Walking trials were conducted on a linear walkway (1.5m×15.5m) embedded with two force plates (BERTEC # K80102, Type 45550-08, Bertec Corporation, OH 43212, USA). Kinetic data from force plates were measured at a sampling rate of 1200 Hz. The walkway was covered with baseline vinyl title (Armstrong). Floor surface was covered by slippery mixture (soap/water = 2/3) to reduce the surface coefficient of friction (COF) (dynamic COF was tested to be 0.07). Uniform experimental shoes were provided to participants to minimize shoe sole differences.

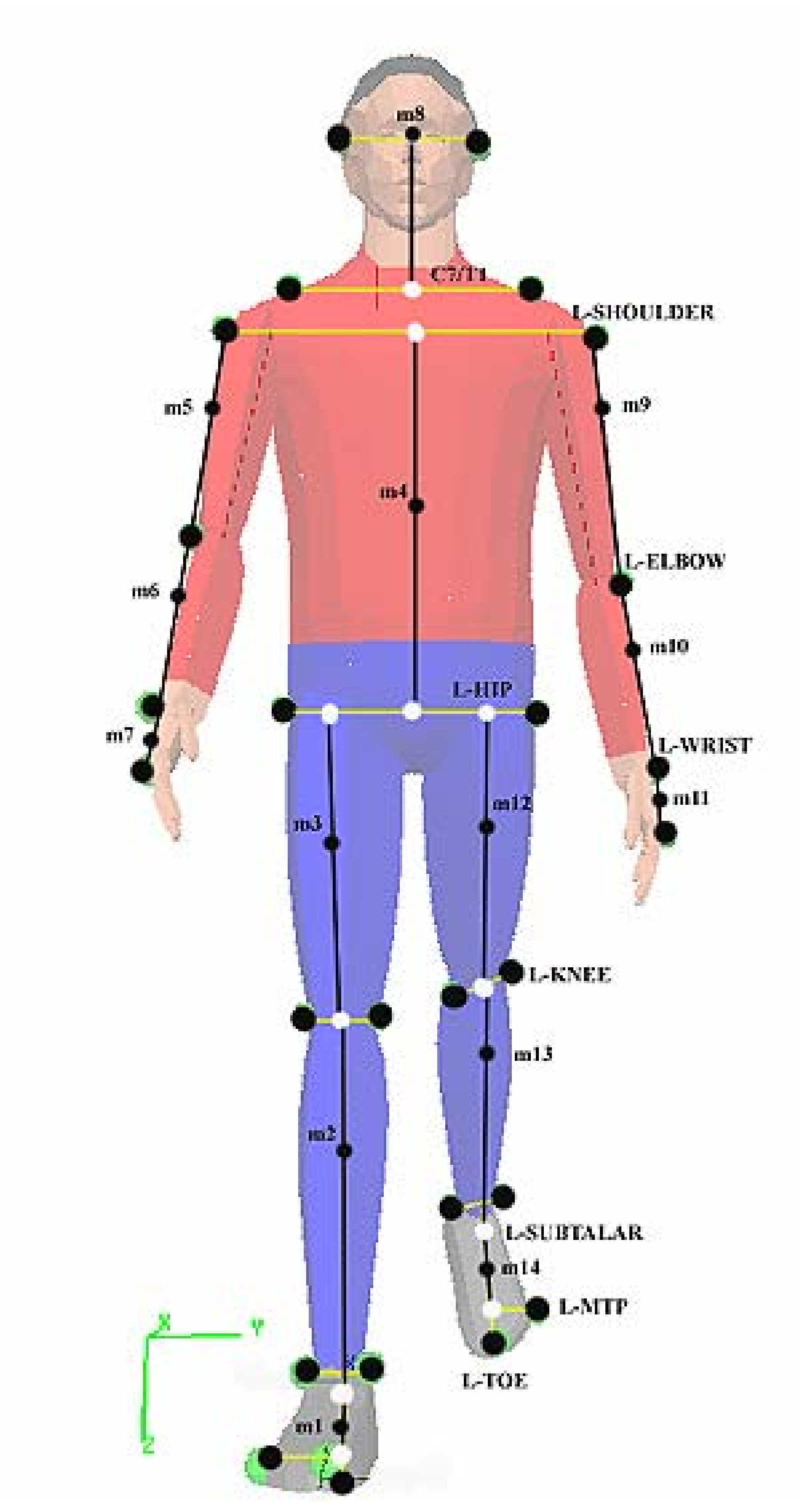

Figure 3.2-1 Marker configuration and internal landmarks illustrated in the frontal plane

Twenty-six small spherical reflective markers were placed over anatomical landmarks of participants according to the marker configuration described by Lockhart et al (2002). The marker configuration of this whole body model was illustrated in Figure 3.2-1. A six-camera ProReflex system (Qualysis) was used to collect three-dimensional position data of the participant while walking. A fall-arresting rig was used to prevent participants from hitting the ground except their feet (Lockhart et al., 2002). Kinematic data were sampled and recorded at 120 Hz.

BioDex dynamometer (Biodex Medical Systems, Inc., Shirley, NY) was used to collect isokinetic strength at the participant's ankle and knee joints.

3.2.2. Procedure

Participants were involved in two separate sessions within one week. The fall-arresting system and walking instructions were introduced to each participant during the initial familiarization session. Reflective markers were then attached to the anatomical landmarks.

During the experiment, participants were instructed to walk naturally across a dry vinyl floor surface for 20 minutes. Data of normal walking trials were then collected. Within a subsequent 20 minute session, a slippery surface was introduced without participants' awareness and kinetic data from two force plates and kinematic data of reflective markers were collected simultaneously.

Within one week after the walking trial, participants were scheduled to measure isokinetic strength at ankle and knee joints. Each participant was allowed to walk on the track for 2-3 minutes to warm up lower extremity muscles before data collection. Dynamic maximum ankle plantarflexor/dorsiflexor and knee extensor/flexor torque measurements at three exertion speed levels ($30^\circ/s$, $60^\circ/s$ and $120^\circ/s$) were recorded randomly.

3.3. Experimental variables

3.3.1. Independent variables

Age groups (Between subjects). There are two age groups, younger group (19-35 years old) and older group (65-89 years old), for current study.

Walking conditions (Within subjects). Two types of walking conditions, normal walking (on dry surface) and reactive-recovery (on slippery surface), are considered as independent variables.

3.3.2. Dependent variables

Lower extremity joint moments were determined through inverse dynamics approach during stance phase where the target foot contacts the contaminated slippery surface. Resultant joint moments were normalized by participant's weight to minimize confounding effect of body mass (Moisio, Sumner, Shott, & Hurwitz, 2003).

Reactive joint moment activation time (JMA Time)

The reactive joint moment activation time (JMA Time) was defined as the duration from heel contact to the activation of reactive joint moment. The threshold of heel contact was defined as the instant when the vertical ground reaction force (GRF) exceeded 7 N. Slip start was defined as the instant when the first minimum heel velocity after the heel contact occurred. Slip end was defined as the instant when the first maximum horizontal heel velocity occurred. The threshold of reactive joint moment activation was then defined as the average joint moment plus 2 SD (standard deviation) during the interval from slip start to slip end (Figure 3.3-1).

The JMA Time was used to characterize the time needed to initiate reactive joint moment responding to slip-induced perturbations.

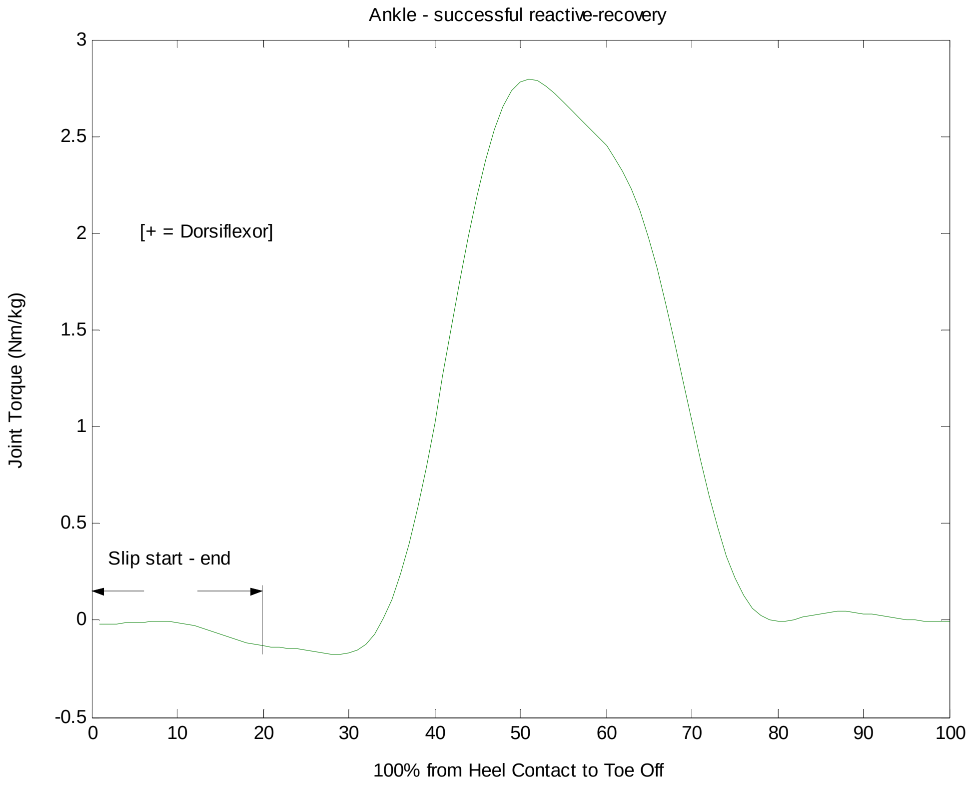

Figure 3.3-1 Typical ankle sagittal plane joint moment profile during reactive-recovery. The period from slip start to slip end will be used to determine joint moment activation threshold.

Joint moment activation to peak time (JMAP Time)

The joint moment activation to peak time (JMAP Time) was defined as the duration from the joint moment activation to the peak joint moment. Time of peak joint moment was defined as the instant when the maximum joint moment magnitude occurred during stance phase.

The JMAP Time was used to characterize the time needed to generate maximum joint moment in order to achieve successful reactive-recovery after reactive joint moment was initiated.

Peak joint moment magnitude (JMP Magnitude)

The peak joint moment magnitude (JMP Magnitude) was defined as the magnitude of the maximum joint moment during stance phase. This variable was used to characterize joint moment generation differences between age groups during reactive-recovery.

Joint moment generation ratio (JMG Ratio)

The joint moment generation ratio (JMG Ratio) was defined as the peak sagittal joint moment magnitude in ankle and knee joints to the maximum isokinetic muscle torque measurement at that joint. The strength measurement at exertion level of $120^{\circ}/s$ was used based on the instant ankle sagittal angular velocity at heel contact. JMG Ratio was used to characterize the ability of moment generation during reactive-recovery.

Peak joint moment ratio (JMP Ratio)

The peak joint moment ratio (JMP Ratio) for each plane was defined as the percentage of the peak joint moment in that plane relative to the summation of the peak joint moments in all three reference planes for specific joint. Consequently, for each lower extremity joint (ankle, knee or hip), three specific (plane $\times$joint) JMP Ratios were determined.

The JMP Ratio was used to characterize the distribution of peak joint moments on each of the three reference planes for specific joint and the relative contribution of each reference plane in terms of successful reactive-recovery.

Joint moment generation rate (JMG Rate)

The joint moment generation rate (JMG Rate) for each plane at each joint was defined as the slope of peak joint moment in that plane relative to the corresponding JMAP Time.

Peak joint power magnitude (JMP Power)

The peak joint power magnitude (JMP Power) at each joint was determined as maximum joint power after slip end. Joint power was computed as the dot product of joint angular velocity and net joint moment during stance phase (Riley et al., 2001).

3.4. Inverse dynamics

3.4.1. Input Information and Data Preprocessing

Initial input information for inverse dynamics included direct measurements from experimental setting and estimates from anthropometric distributions (Dempster, Gabel, & Felts, 1959; Webb Associates, 1978). There were two types of direct measurements, the 3D coordinates of reflective markers measured from infrared camera system and the ground reaction force and moment measurements from force plates. Segment mass and inertial properties were estimated using existing anthropometric statistical distributions. Data preprocessing included filtering raw data and coordinate synchronization between different measurements (i.e. kinematic measurement and force-plate measurement).

Coordinates of external landmarks

Coordinates of external landmarks were directly generated by QTrac System which capture and reconstruct three-dimensional (3D) locations of reflective markers using six infrared cameras. As illustrated in Figure 3.2-1， there were sixteen markers, which were used in the proposed study, placed on the anatomical landmarks of the lower extremity. Specifically, the sixteen landmarks represented toe, lateral and medial ankle, heel, lateral surface, lateral and medial knee, and hip for right and left sides.

The reference frame of the landmark coordinates was consistent with global (lab fixed) reference frame. Specifically, X axis represented anterior-posterior walking direction; Y axis represented medial-lateral direction; Z axis represented vertical direction.

Coordinates of external landmarks were digitally low-pass filtered by a zero-lag fourth order Butterworth filter with a cut-off frequency of 6 Hz.

External kinetic quantities and point of application

Raw force-plate data exported from LabView, included three scalar forces ($F_{X,GRF}$, $F_{Y,GRF}$, $F_{Y,GRF}$) and three scalar moments ($M_{X,GRF}$, $M_{Y,GRF}$, $M_{Z,GRF}$). These measurements were digitally low-pass filtered by a zero-phase fourth order Butterworth filter with a cut-off frequency of 12 Hz.

Environmental reactions to human body during level walking included ground reaction forces and moments. The point of application of the force (Center Of Pressure) and the couple (torque) were calculated from the reaction force and moment components in equations(3.3)(3.4)(3.5):

Table 3.4-1 Segment weight percentages

Segment weights	
Segment	Mean segment weight as percentage of total body weight
Right Thigh	9.6
Left Thigh	9.7
Right Shank	4.5
Left Shank	4.5
Left Foot	1.4
Right Foot	1.4

$$P_{X,COP} = \frac{-h \cdot F_{X,GRF} - M_{Y,GRF}}{F_{Z,GRF}} \tag{3.3}$$

$$P_{Y,COP} = \frac{-h \cdot F_{Y,GRF} + M_{X,GRF}}{F_{Z,GRF}} \tag{3.4}$$

$$T_{Z,GRF} = M_{Z,GRF} - P_{X,COP} \cdot F_{Y,GRF} + P_{Y,COP} \cdot F_{X,GRF} \tag{3.5}$$

Where, $P_{X,COP}$ and $P_{Y,COP}$ are the coordinates of the point of application of the GRF; *h* is the thickness, above the force plate top surface, of the vinyl tile covering the force plate; and $\vec{T}_{GRF}$, $\vec{T}_{GRF} = \begin{pmatrix} 0 & 0 & T_{Z,GRF} \end{pmatrix}$, is the couple acting on the force plate.

Body Segment Parameters

Body Segment Parameters (BSPs) used as input for proposed inverse dynamics included lower extremity segment mass and inertial properties. Available individual anthropometric measurements included participants' height and weight. Segment mass then was estimated as a percentage of whole body weight from Table 3.4-1, which was adapted from Dempster et al., 1959. Segment inertia properties were obtained from the population distribution, Table 3.4-2, adapted from Webb Associates, 1978.

Table 3.4-2 Segment moments of inertia

Segment moments of inertia ($kg \cdot cm^2$)				
		Small: 5%	Medium: 50%	Large: 95%
		168.2cm / 65.2kg	178.4cm / 81.5kg	188.6cm / 97.7kg
Thigh	I_{ap}	1123.9	1666.6	2334.9
	I_{ml}	1164.9	1726.8	2419.5
	$I_{\lg}$	214.9	318.7	446.5
Shank	I_{ap}	369.3	534.3	736.9
	I_{ml}	369.3	534.3	736.9
	$I_{\lg}$	26.8	38.8	53.5
Foot	I_{ap}	40.1	52.1	66.6
	I_{ml}	36.5	47.4	60.6
	$I_{\lg}$	8.8	11.4	14.5

I_{ap}, I_{ml} and $I_{\lg}$ are the moments of inertia about the anteroposterior, mediolateral, and longitudinal axes of the segments, respectively.

3.4.2. Determination of Linear Quantity

Determine joint center location

Joint center locations were determined through external landmarks. All the percentage data used in this section were adapted from the previously published biomechanical model (Lockhart et al., 2000). Specifically, considering the right side, the ankle joint center was defined as the midpoint of lateral malleolus and medial malleolus on one side.

$$\vec{P}_{ankle} = \frac{\left(\vec{P}_{l\_malleolus} + \vec{P}_{m\_malleolus}\right)}{2} \tag{3.6}$$

where $\vec{P}_{ankle}, \vec{P}_{l\_malleolus}, \vec{P}_{m\_malleolus}$ are the position vectors of ankle joint center, lateral malleolus marker, and medial malleolus marker, respectively.

The knee joint center was defined as the midpoint of lateral condyle and medial condyle on one side.

$$\vec{P}_{knee} = \frac{\left(\vec{P}_{l\_condyle} + \vec{P}_{m\_condyle}\right)}{2} \tag{3.7}$$

where $\vec{P}_{knee}, \vec{P}_{l\_condyle}, \vec{P}_{m\_condyle}$ are the position vectors of knee joint center, lateral condyle marker, and medial condyle marker, respectively.

The hip joint center was defined as the medially 19.7% of the distance between the right and left greater trochanter markers.

$$\vec{P}_{hip} = 19.7\% \cdot \left(\vec{P}_{l\_trochanter} - \vec{P}_{r\_trochanter}\right) + \vec{P}_{r\_trochanter} \tag{3.8}$$

where $\vec{P}_{hip}, \vec{P}_{l\_trochanter}, \vec{P}_{r\_trochater}$ are the position vectors of hip joint center, left trochanter marker, and right trochanter marker, respectively.

Determine segment COM location

Segment center of mass (COM) locations were determined through combination of external landmarks and joint center locations. Specifically, considering the right side, the foot COM was defined as the midpoint of ankle joint center and the metatarsal head II.

$$\vec{P}_{foot} = \frac{\left(\vec{P}_{ankle} + \vec{P}_{toe}\right)}{2} \tag{3.9}$$

where $\vec{P}_{foot}, \vec{P}_{toe}$ are the position vectors of foot COM, and metatarsal head II marker, respectively.

The shank COM was defined as the 43.3% below knee joint center in the line connecting knee and ankle joints.

$$\vec{P}_{shank} = 43.3\% \cdot \left(\vec{P}_{ankle} - \vec{P}_{knee}\right) + \vec{P}_{knee} \tag{3.10}$$

where $\vec{P}_{shank}$ is the position vectors of shank COM.

The thigh COM was defined as the 43.3% below hip joint center in the line connecting hip and knee joints.

$$\vec{P}_{thigh} = 43.3\% \cdot \left(\vec{P}_{knee} - \vec{P}_{hip}\right) + \vec{P}_{hip} \tag{3.11}$$

where $\vec{P}_{thigh}$ is the position vectors of thigh COM.

Determine linear velocity and accelerations

Linear velocity, which is the first time-derivative of position vector (denoted as $\dot{\vec{P}}$), and acceleration, which is the second time-derivative of position vector (denoted as $\ddot{\vec{P}}$), of segment COM was determined using central difference method. For example, considering foot COM linear velocity and acceleration at given time *t*,

$$\dot{\vec{P}}^{t}_{foot} = \frac{\left(\vec{P}^{t+1}_{foot} - \vec{P}^{t-1}_{foot}\right)}{2 \cdot \Delta t} \tag{3.12}$$

$$\ddot{\vec{P}}^{t}_{foot} = \frac{\left(\vec{P}^{t+2}_{foot} + \vec{P}^{t-2}_{foot} - 2 \cdot \vec{P}^{t}_{foot}\right)}{\Delta t^2} \tag{3.13}$$

where Δt is the time interval between two continuous data samples, while $\vec{P}^{t-2}_{foot}, \vec{P}^{t-1}_{foot}, \vec{P}^{t}_{foot}, \vec{P}^{t+1}_{foot}, \vec{P}^{t+2}_{foot}$ are the position vectors of foot COM at time *t-2*, *t-1*, *t*, *t+1*, *t+2*, respectively.

3.4.3. Coordinate Systems Transformation

A local (segmental) coordinate system is needed to define the Euler angles of a segment in space, to determine the transformation matrix between different coordinate systems, and to express the resultant vector quantity (3D joint moment in current study) in space. A local coordinate system for each joint in the lower extremity was defined using available position vectors and Gram-Schmidt orthogonalization algorithm.

Given an arbitrary basis $\{\alpha_1, \alpha_2, \alpha_3\}$ for a 3-dimensional inner product space *V*, the Gram-Schmidt orthogonalization process (Bradley, 1975) constructs an orthogonal basis $\{\gamma_1, \gamma_2, \gamma_3\}$ for *V*. Two vectors γ_2 and γ_3, which are both orthogonal to the γ_1, are computed as:

$$\gamma_2 = \alpha_2 - \frac{\langle \alpha_2, \gamma_1 \rangle}{\|\gamma_1\|} \tag{3.14}$$

$$\gamma_3 = \alpha_3 - \frac{\langle \alpha_3, \gamma_1 \rangle}{\|\gamma_1\|^2}\gamma_1 - \frac{\langle \alpha_3, \gamma_2 \rangle}{\|\gamma_2\|^2}\gamma_2 \tag{3.15}$$

Referring to the muscular-skeletal model, for each segment, at least two non-parallel vectors approximately located in frontal plane can be established using 3 non-collinear external markers. Furthermore, three principle axes $\{\gamma_1, \gamma_2, \gamma_3\}$ for each segment can be derived consequently, where γ_1 represents the longitudinal axis and the vectors α_2 and α_3 are defined as those two vectors that are not scalar multiples of γ_1, γ_2 or γ_3.

Local joint coordinate system

For the right ankle joint, local y axis (denoted as $\vec{u}_{ankle}$) was defined as the vector pointing from lateral malleolus to medial malleolus; local y-z vector (representing vector in y-z plane) was defined as the vector pointing from surface marker (located on the shoe surface below lateral malleolus marker perpendicularly when the participant was in standing posture) to ankle joint center. Local x axis (denoted as $\vec{v}_{ankle}$) was then defined as the cross product of y axis and y-z vector. Thus, local z axis (denoted as $\vec{w}_{ankle}$) was defined as the cross product of local x axis and local y axis. The unit vector triad $\left(\vec{u}_{ankle} \quad \vec{v}_{ankle} \quad \vec{w}_{ankle}\right)^T$ for the right ankle was derived in Eq.(3.16)(3.17)(3.18):

$$\vec{u}_{ankle} = \frac{\left(\vec{P}_{m\_malleolus} - \vec{P}_{l\_malleolus}\right)}{\left|\vec{P}_{m\_malleolus} - \vec{P}_{l\_malleolus}\right|} \tag{3.16}$$

$$\vec{v}_{ankle} = \frac{\vec{u}_{ankle} \times \left(\vec{P}_{ankle} - \vec{P}_{surface}\right)}{\left|\vec{u}_{ankle} \times \left(\vec{P}_{ankle} - \vec{P}_{surface}\right)\right|} \tag{3.17}$$

$$\vec{w}_{ankle} = \frac{\vec{v}_{ankle} \times \vec{u}_{ankle}}{\left|\vec{v}_{ankle} \times \vec{u}_{ankle}\right|} \tag{3.18}$$

For the right knee joint, local y axis (denoted as $\vec{u}_{knee}$) was defined as the vector pointing from lateral femoral condyle to medial femoral condyle; local y-z vector was defined as the vector pointing from ankle joint center to knee joint center. Local x axis (denoted as $\vec{v}_{knee}$) was then defined as the cross product of y axis and y-z vector. Thus, local z axis (denoted as $\vec{w}_{knee}$) was defined as the cross product of x axis and y axis. The unit vector triad $\left(\vec{u}_{knee} \quad \vec{v}_{knee} \quad \vec{w}_{knee}\right)^T$ for the right knee joint was expressed as follows:

$$\vec{u}_{knee} = \frac{\left(\vec{P}_{m\_condyle} - \vec{P}_{l\_condyle}\right)}{\left|\vec{P}_{m\_condyle} - \vec{P}_{l\_condyle}\right|} \tag{3.19}$$

$$\vec{v}_{knee} = \frac{\vec{u}_{knee} \times \left(\vec{P}_{knee} - \vec{P}_{ankle}\right)}{\left|\vec{u}_{knee} \times \left(\vec{P}_{knee} - \vec{P}_{ankle}\right)\right|} \tag{3.20}$$

$$\vec{w}_{knee} = \frac{\vec{v}_{knee} \times \vec{u}_{knee}}{\left|\vec{v}_{knee} \times \vec{u}_{knee}\right|} \tag{3.21}$$

For the right hip joint, local y axis (denoted as $\vec{u}_{hip}$) was defined as the vector pointing from right greater trochanter to left greater trochanter; local y-z vector was defined as the vector pointing from knee joint center to hip joint center. Local x axis (denoted as $\vec{v}_{hip}$) was then defined as the cross product of y axis and y-z vector. Thus, local z axis (denoted as $\vec{w}_{hip}$) was defined as the cross product of x axis and y axis. The unit vector triad $\left(\vec{u}_{hip} \quad \vec{v}_{hip} \quad \vec{w}_{hip}\right)^T$ for the right hip joint was derived in Eq.(3.22)(3.23)(3.24):

$$\vec{u}_{hip} = \frac{\left(\vec{P}_{l\_trochanter} - \vec{P}_{r\_trochanter}\right)}{\left|\vec{P}_{l\_trochanter} - \vec{P}_{r\_trochanter}\right|} \tag{3.22}$$

$$\vec{v}_{hip} = \frac{\vec{u}_{hip} \times \left(\vec{P}_{hip} - \vec{P}_{knee}\right)}{\left|\vec{u}_{hip} \times \left(\vec{P}_{hip} - \vec{P}_{knee}\right)\right|} \tag{3.23}$$

$$\vec{w}_{hip} = \frac{\vec{v}_{hip} \times \vec{u}_{hip}}{\left|\vec{v}_{hip} \times \vec{u}_{hip}\right|} \tag{3.24}$$

Local segmental coordinate system

In this proposed study, a local segmental coordinate system for each segment was defined as the same with local joint coordinate system for corresponding proximal joints, because the defined unit vector triad for each joint was actually free vectors. Namely, right foot local coordinate was the same as the right ankle joint ($\vec{u}_{foot} = \vec{u}_{ankle}; \vec{w}_{foot} = \vec{w}_{ankle}; \vec{w}_{foot} = \vec{w}_{ankle}$); right shank local coordinate was the same as the right knee joint ($\vec{u}_{shank} = \vec{u}_{knee}; \vec{w}_{shank} = \vec{w}_{knee}; \vec{w}_{shank} = \vec{w}_{knee}$); right hip local coordinate was the same as the right hip ($\vec{u}_{thigh} = \vec{u}_{hip}; \vec{w}_{thigh} = \vec{w}_{hip}; \vec{w}_{thigh} = \vec{w}_{hip}$).

Transformation matrix

The transformation matrix (denoted as *T*) from the global coordinate system to a particular local coordinate system is defined as:

$$T_{L/G} = \begin{pmatrix} \vec{i} \cdot \vec{i}' & \vec{j} \cdot \vec{i}' & \vec{k} \cdot \vec{i}' \\ \vec{i} \cdot \vec{j}' & \vec{j} \cdot \vec{j}' & \vec{k} \cdot \vec{j}' \\ \vec{i} \cdot \vec{k}' & \vec{j} \cdot \vec{k}' & \vec{k} \cdot \vec{k}' \end{pmatrix} \tag{3.25}$$

where $\begin{pmatrix} \vec{i} & \vec{j} & \vec{k} \end{pmatrix}$ is the unit vectors of the global coordinate system, and $\begin{pmatrix} \vec{i}' & \vec{j}' & \vec{k}' \end{pmatrix}$ is the unit vector of a particular local coordinate system.

Similarly, the inverse transformation matrix ($T_{L/G}$) is also defined as

$$T_{G/L} = \begin{pmatrix} \vec{i} \cdot \vec{i}' & \vec{i} \cdot \vec{j}' & \vec{i} \cdot \vec{k}' \\ \vec{j} \cdot \vec{i}' & \vec{j} \cdot \vec{j}' & \vec{j} \cdot \vec{k}' \\ \vec{k} \cdot \vec{i}' & \vec{k} \cdot \vec{j}' & \vec{k} \cdot \vec{k}' \end{pmatrix} \tag{3.26}$$

Similarly, $T_{B/A}$, the transformation matrix from a particular local coordinate system *A* to another local coordinate system *B* is derived as

$$T_{B/A} = \begin{pmatrix} \vec{i}_A \cdot \vec{i}_B{}' & \vec{j}_A \cdot \vec{i}_B{}' & \vec{k}_A \cdot \vec{i}_B{}' \\ \vec{i}_A \cdot \vec{j}_B{}' & \vec{j}_A \cdot \vec{j}_B{}' & \vec{k}_A \cdot \vec{j}_B{}' \\ \vec{i}_A \cdot \vec{k}_B{}' & \vec{j}_A \cdot \vec{k}_B{}' & \vec{k}_A \cdot \vec{k}_B{}' \end{pmatrix} \tag{3.27}$$

Therefore, the transformation matrix from global (lab-fixed) coordinate to right foot coordinate system was derived as

$$T_{foot/G} = \begin{pmatrix} \vec{i} \cdot \vec{u}_{foot} & \vec{j} \cdot \vec{u}_{foot} & \vec{k} \cdot \vec{u}_{foot} \\ \vec{i} \cdot \vec{v}_{foot} & \vec{j} \cdot \vec{v}_{foot} & \vec{k} \cdot \vec{v}_{foot} \\ \vec{i} \cdot \vec{w}_{foot} & \vec{j} \cdot \vec{w}_{foot} & \vec{k} \cdot \vec{w}_{foot} \end{pmatrix} \tag{3.28}$$

where $\left(\vec{u}_{foot} \quad \vec{v}_{foot} \quad \vec{w}_{foot} \right)$ is the unit vectors of right foot coordinate system.

The GRF vector transformed into right foot coordinate system was derived as

$$\vec{F}_{GRF}^{\ (foot)} = T_{foot/G} \cdot \vec{F}_{GRF} \tag{3.29}$$

where $\vec{F}_{GRF}$ is the GRF expressed in global coordinate system, and $\vec{F}_{GRF}^{\ (foot)}$ is the GRF expressed in right foot coordinate system.

3.4.4. Determination of Angular Quantity

Euler angles and Rotation Matrix

Euler angles are a new set of three coordinates by which the orientation of a rigid body can be specified in space. Euler angles can further be used to derive the rotational velocity and acceleration of a rigid body.

Considering a particular local *xyz* coordinate system which is initially coincident with a global *XYZ* coordinate system, a series of three rotations about the body axes, performed in the proper sequence, is sufficient to allow the xyz system to reach any orientation in space. The three rotations (illustrated in Figure) defined in current study were:

1. A positive rotation ϕ about the X axis, resulting in the primed system.

2. A positive rotation θ about the Y' axis, resulting in the double-primed system.

3. A positive rotation ψ about the Z'' axis, resulting in the final unprimed system.

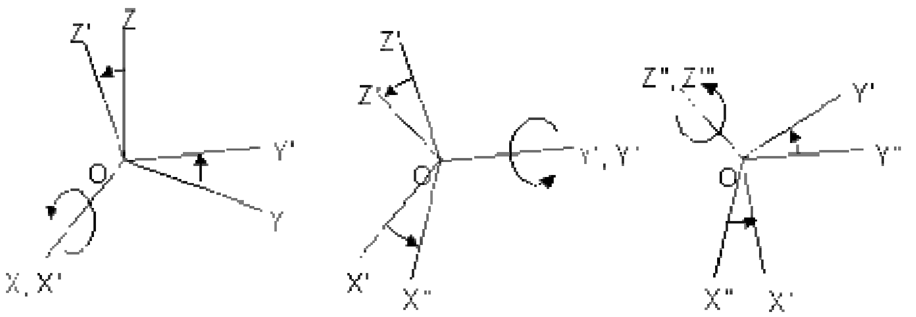

Figure 3.4-1 Illustrate three successive rotations to change the orientation of a particular local coordinate system from the global coordinate system to the $X'''Y'''Z'''$ system, which is the final xyz system. The intermediate coordinate systems are denoted as $X'Y'Z'$ system and $X''Y''Z''$ system, respectively.

Equations that indicate the rotations were derived as

$$\begin{pmatrix} X' \\ Y' \\ Z' \end{pmatrix} = \begin{pmatrix} 1 & 0 & 0 \\ 0 & \cos\phi & \sin\phi \\ 0 & -\sin\phi & \cos\phi \end{pmatrix} \begin{pmatrix} X \\ Y \\ Z \end{pmatrix} \tag{3.30}$$

$$\begin{pmatrix} X'' \\ Y'' \\ Z'' \end{pmatrix} = \begin{pmatrix} \cos\theta & 0 & -\sin\theta \\ 0 & 1 & 0 \\ \sin\theta & 0 & \cos\theta \end{pmatrix} \begin{pmatrix} X' \\ Y' \\ Z' \end{pmatrix} \tag{3.31}$$

$$\begin{pmatrix} X''' \\ Y''' \\ Z''' \end{pmatrix} = \begin{pmatrix} \cos\psi & \sin\psi & 0 \\ -\sin\psi & \cos\psi & 0 \\ 0 & 0 & 1 \end{pmatrix} \begin{pmatrix} X'' \\ Y'' \\ Z'' \end{pmatrix} \tag{3.32}$$

The successive rotations from coordinate system A to coordinate system B are expressed as multiplication of the rotation matrices

$$\begin{pmatrix} x \\ y \\ z \end{pmatrix} = R_Z(\psi) \cdot R_Y(\theta) \cdot R_X(\phi) \cdot \begin{pmatrix} X \\ Y \\ Z \end{pmatrix} = T_{B/A} \cdot \begin{pmatrix} X \\ Y \\ Z \end{pmatrix} \tag{3.33}$$

where transformation matrix $T_{B/A}$ are expressed in terms of the successive rotation angles:

$$T_{B/A} = \begin{pmatrix} \cos\theta \cdot \cos\psi & \begin{matrix} \sin\phi \cdot \sin\theta \cdot \cos\psi \\ +\cos\phi \cdot \sin\psi \end{matrix} & \begin{matrix} -\cos\phi \cdot \sin\theta \cdot \cos\psi \\ +\sin\phi \cdot \sin\psi \end{matrix} \\ -\cos\theta \cdot \sin\psi & \begin{matrix} -\sin\phi \cdot \sin\theta \cdot \sin\psi \\ +\cos\phi \cdot \cos\psi \end{matrix} & \begin{matrix} \cos\phi \cdot \sin\theta \cdot \sin\psi \\ +\sin\phi \cdot \cos\psi \end{matrix} \\ \sin\theta & -\sin\phi \cos\theta & \cos\phi \cdot \cos\theta \end{pmatrix} \tag{3.34}$$

Determination of Euler angles

As stated in Eq.(3.27), a particular transformation matrix ($T_{B/A}$) can be determined from a particular segment local coordinate system to global coordinate system.

$$T_{B/A} = \begin{pmatrix} t_{11} & t_{12} & t_{13} \\ t_{21} & t_{22} & t_{23} \\ t_{31} & t_{32} & t_{33} \end{pmatrix} \tag{3.35}$$

Derived from equation (3.35) and(3.34),

$$\begin{pmatrix} t_{11} & t_{12} & t_{13} \\ t_{21} & t_{22} & t_{23} \\ t_{31} & t_{32} & t_{33} \end{pmatrix} = \begin{pmatrix} \cos\theta \cdot \cos\psi & \begin{matrix} \sin\phi \cdot \sin\theta \cdot \cos\psi \\ +\cos\phi \cdot \sin\psi \end{matrix} & \begin{matrix} -\cos\phi \cdot \sin\theta \cdot \cos\psi \\ +\sin\phi \cdot \sin\psi \end{matrix} \\ -\cos\theta \cdot \sin\psi & \begin{matrix} -\sin\phi \cdot \sin\theta \cdot \sin\psi \\ +\cos\phi \cdot \cos\psi \end{matrix} & \begin{matrix} \cos\phi \cdot \sin\theta \cdot \sin\psi \\ +\sin\phi \cdot \cos\psi \end{matrix} \\ \sin\theta & -\sin\phi \cos\theta & \cos\phi \cdot \cos\theta \end{pmatrix} \tag{3.36}$$

Therefore, the Euler angles ϕ , θ , ψ will be determined based on following expressions.

$$\theta = \{ \begin{matrix} \sin^{-1}(t_{31}) \\ or \\ \pi - \sin^{-1}(t_{31}) \end{matrix} \tag{3.37}$$

and, if $\theta \neq \pm\frac{\pi}{2}$

$$\phi = \begin{cases} \tan^{-1}\left[\frac{-t_{32}}{t_{33}}\right] & (t_{33} \cdot \cos\theta > 0) \\ \tan^{-1}\left[\frac{-t_{32}}{t_{33}}\right] + \pi & (t_{33} \cdot \cos\theta < 0) \end{cases} \tag{3.38}$$

$$\psi = \begin{cases} \tan^{-1}\left[\frac{-t_{21}}{t_{11}}\right] & (t_{11} \cdot \cos\theta > 0) \\ \tan^{-1}\left[\frac{-t_{21}}{t_{11}}\right] + \pi & (t_{11} \cdot \cos\theta < 0) \end{cases} \tag{3.39}$$

In this way, the Euler angles that specify the orientation of a particular segment can be determined in space. With consideration of the continuous angle history, specific Euler angle time histories for each segment in current study were then determined.

Specifically, for the foot segment, θ is the Euler angle about local *x* axis, which is approximately equivalent to the anteroposterior axis of foot. ϕ is the Euler angle about local *y* axis, which is approximately equivalent to the medial-lateral axis of foot. ψ is the Euler angle about local *z* axis, which is equivalent to the longitudinal axis of foot.

Similarly, for the shank segment, θ is the Euler angle about local *x* axis, which is approximately equivalent to the anteroposterior axis of shank. ϕ is the Euler angle about local *y* axis, which is approximately equivalent to the medial-lateral axis of shank. ψ is the Euler angle about local *z* axis, which is equivalent to the longitudinal axis of shank.

For the thigh segment, θ is the Euler angle about local *x* axis, which is approximately equivalent to the anteroposterior axis of thigh. ϕ is the Euler angle about local *y* axis, which is approximately equivalent to the medial-lateral axis of thigh. ψ is the Euler angle about local *z* axis, which is equivalent to the longitudinal axis of thigh.

Determination of angular velocity and acceleration

Segmental angular velocity was computed by taking the first time-derivative of obtained Euler angles for a particular segment and coordinate transformation.

After taking the first time-derivative of ϕ, θ, ψ, three independent angular vectors is obtained though their directions are not in one common reference coordinate system. As shown in Figure 3.4-1, the direction of the first rotation is about the X / X' axis; the second rotation, is about the Y' / Y'' axis and the third rotation, is about Z'' / Z''' axis.

The angular velocity vectors in one common reference coordinate system can be obtained with assistance of rotation matrix. Specifically, the angular velocity vector ($\vec{\omega}_{B/A}^{(B)}$) of coordinate system *B* rotated from coordinate system *A*, expressed in coordinate system *B*, is derived as

$$\begin{aligned}\vec{\omega}_{B/A}^{(B)} &= R_Z(\psi)\cdot R_Y(\theta)\cdot\begin{pmatrix}\dot{\phi}\\0\\0\end{pmatrix} + R_Z(\psi)\cdot\begin{pmatrix}0\\\dot{\theta}\\0\end{pmatrix} + \begin{pmatrix}0\\0\\\dot{\psi}\end{pmatrix}\\ &= \begin{pmatrix}\cos\theta\cdot\cos\psi & \sin\psi & 0\\ -\cos\theta\cdot\sin\psi & \cos\psi & 0\\ \sin\theta & 0 & 1\end{pmatrix}\begin{pmatrix}\dot{\phi}\\\dot{\theta}\\\dot{\psi}\end{pmatrix}\end{aligned} \tag{3.40}$$

Similarly, this angular velocity vector ($\vec{\omega}_{B/A}^{(A)}$) expressed in coordinate system *A*, is derived as

$$\begin{aligned}\vec{\omega}_{B/A}^{(A)} &= \begin{pmatrix}\dot{\phi}\\0\\0\end{pmatrix} + R_X(-\phi)\cdot\begin{pmatrix}0\\\dot{\theta}\\0\end{pmatrix} + R_X(-\phi)\cdot R_Y(-\theta)\cdot\begin{pmatrix}0\\0\\\dot{\psi}\end{pmatrix}\\ &= \begin{pmatrix}1 & 0 & \sin\theta\\ 0 & \cos\phi & -\sin\phi\cdot\cos\theta\\ 0 & \sin\phi & \cos\phi\cdot\cos\theta\end{pmatrix}\begin{pmatrix}\dot{\phi}\\\dot{\theta}\\\dot{\psi}\end{pmatrix}\end{aligned} \tag{3.41}$$

After the angular velocity vector is expressed in one common coordinate system, the angular acceleration vector can be simply determined by taking the first time-derivative of the angular velocity.

Specifically, for the right foot segment, the local coordinate system was first defined via Eq. (3.16)(3.17)(3.18). The transformation matrix from global coordinate system to local coordinate system was then derived via Eq. (3.28). From Eq. (3.36) and Eq.(3.37)(3.38)(3.39), the time history of Euler angles of right foot segment was computed. Angular velocity of right foot expressed in global system was then determined using Eq.(3.40)(3.41), while angular acceleration of right foot expressed in global system $\ddot{\vec{\omega}}_{foot}$ was determined by taking the first derivative of $\dot{\vec{\omega}}_{foot}$.

Determination of angular momentum

The first time-derivative of angular momentum is needed in computing joint moment using rotational equation of motion(3.47). The equations are derived as matrix form Eq. (3.42)as well as scalar form Eq.(3.43)(3.44)(3.45):

$$\begin{aligned}
\dot{\vec{H}} &= \frac{d\vec{H}}{dt} \\
&= \left(\frac{d\vec{H}}{dt}\right)^{(local)} + \vec{\omega}\times\vec{H} \\
&= I\cdot\dot{\vec{\omega}} + \vec{\omega}\times(I\cdot\vec{\omega}) \\
&= \begin{pmatrix} I_{XX} & 0 & 0 \\ 0 & I_{YY} & 0 \\ 0 & 0 & I_{ZZ} \end{pmatrix}\begin{pmatrix} \dot{\vec{\omega}}_X \\ \dot{\vec{\omega}}_Z \\ \dot{\vec{\omega}}_Z \end{pmatrix} + \begin{pmatrix} \vec{\omega}_X \\ \vec{\omega}_Y \\ \vec{\omega}_Z \end{pmatrix} \times \left(\begin{pmatrix} I_{XX} & 0 & 0 \\ 0 & I_{YY} & 0 \\ 0 & 0 & I_{ZZ} \end{pmatrix}\begin{pmatrix} \vec{\omega}_X \\ \vec{\omega}_Z \\ \vec{\omega}_Z \end{pmatrix}\right)
\end{aligned} \tag{3.42}$$

$$\dot{H}_X = I_{XX}\dot{\omega}_X + (I_{ZZ} - I_{YY})\omega_Y\omega_Z \tag{3.43}$$

$$\dot{H}_Y = I_{YY}\dot{\omega}_Y + (I_{XX} - I_{ZZ})\omega_Z\omega_X \tag{3.44}$$

$$\dot{H}_Z = I_{ZZ}\dot{\omega}_Z + (I_{YY} - I_{XX})\omega_X\omega_Y \tag{3.45}$$

where $\vec{\omega}$ and $\dot{\vec{\omega}}$ are the angular velocity and acceleration expressed in local coordinate system, respectively. I represents inertial property matrix, while I_{XX} , I_{YY} , I_{ZZ} are the moments of inertia of the particular segment.

The moments of inertia I_{XX} , I_{YY} , I_{ZZ} are assumed to be the principle moments of inertia I_{ap} , I_{ml} , $I_{\lg}$ obtained from the segment inertia property table (Table 3.4-2). Theoretically, it is quite possible that the principal axes of inertia of human body segment do not precisely coincide with the segment local axes constructed using anatomical landmarks in proposed study. However, due to the symmetry of most of body segments in sagittal and frontal planes, principal axes are expected to be very close to the segment local axes (Zatsiorsky, 2002). In the majority of biomechanics research, the principal moments of inertia are assumed with respect to the local axes based upon anatomical landmarks.

3.4.5. Determination of Joint Moment

Equation of Motion

Two Newton's equations of motion are of particular importance in the inverse dynamics approach. The first is the translational equation of motion given by Eq. (3.46)

$$\vec{F} = m \cdot \ddot{\vec{P}}_c \tag{3.46}$$

This equation states that the vector sum of the external forces applied on a system is equal to the total mass of the system times the absolute acceleration of the center of mass. The second equation is the rotational equation of motion given by Eq. (3.47)

$$\vec{M} = \dot{\vec{H}} \tag{3.47}$$

This equation states that the vector sum of applied moment $\vec{M}$ on a system is equal to the rate of change of the angular momentum $\vec{H}$. The reference point for calculating these two quantities is either fixed in the global (fixed) coordinate system or located at the center of mass of the system.

Expanded equation of motion

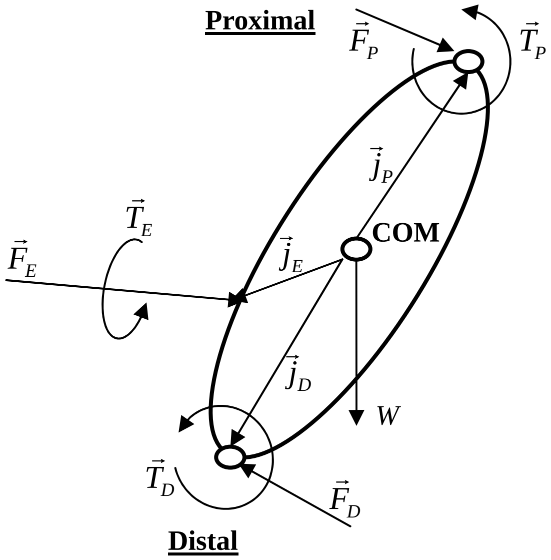

Figure 3.4-2 Typical free body diagram of a body segment

A typical free body diagram of a particular body segment is shown in Figure 3.4-2. The application of the Newton's equations of motion (3.46)(3.47) to a segment shown in the above figure gives the expanded equation of motion

$$\vec{F} = \vec{F}_P + \vec{F}_D + \vec{W} + \vec{F}_E = m \cdot \ddot{\vec{P}}_c \tag{3.48}$$

$$\vec{M} = \vec{j}_P \times \vec{F}_P + \vec{T}_P + \vec{j}_D \times \vec{F}_D + \vec{T}_D + \vec{j}_E \times \vec{F}_E + \vec{T}_E = \dot{\vec{H}} \tag{3.49}$$

Assuming $\vec{F}_D$ and $\vec{T}_D$ are somehow known, these two equations are sufficient to solve for $\vec{F}_P$ and $\vec{T}_P$.

$$\vec{F}_P = m \cdot \ddot{\vec{P}}_c - \vec{F}_D - \vec{W} - \vec{F}_E \tag{3.50}$$

$$\vec{T}_P = \dot{\vec{H}} - \vec{j}_D \times \vec{F}_D - \vec{T}_D - \vec{j}_E \times \vec{F}_E - \vec{T}_E - \vec{j}_P \times \vec{F}_P \tag{3.51}$$

Solution for right foot segment

A free body diagram for the right foot segment was illustrated in Figure 3.4-3. All the related quantities were transformed from global coordinate system to local foot coordinate system. There is no distal segment linked to the foot. Therefore expanded equations of motion (3.50)(3.51) reduced to

$$\vec{F}_{ankle-foot} = m_{foot} \cdot \ddot{\vec{P}}_{foot} - \vec{W}_{foot} - \vec{F}_{GRF} \tag{3.52}$$

$$\vec{T}_{ankle-foot} = \dot{\vec{H}}_{foot} - \vec{j}_{COP} \times \vec{F}_{GRF} - \vec{T}_{GRF} - \vec{j}_{ankle-foot} \times \vec{F}_{ankle-foot} \tag{3.53}$$

where $\vec{F}_{ankle-foot}$ is the net joint force acting on the foot at the ankle joint, $\vec{T}_{ankle-foot}$ is the net joint moment produced by the muscles attached to foot around the ankle joint, and $\vec{j}_{ankle-foot}$ is the relative position vector of the ankle joint to the foot COM.

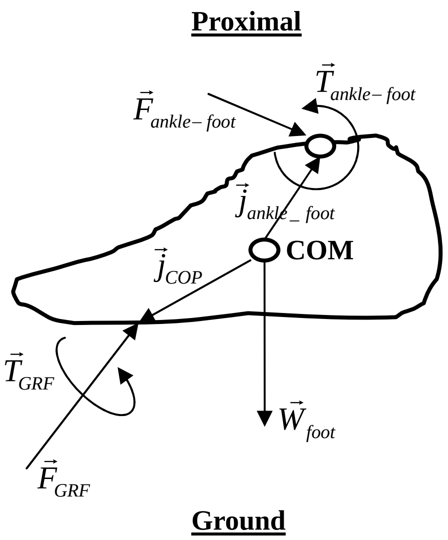

Figure 3.4-3 Free body diagram of foot segment

The resultant joint moment $\vec{T}_{ankle-foot}$ was reported in local foot coordinate system. In addition, to facilitate further computation of the knee joint moment, this ankle joint moment $\vec{T}_{ankle-foot}$ was transformed into local shank coordinate system.

Solution for the right shank segment

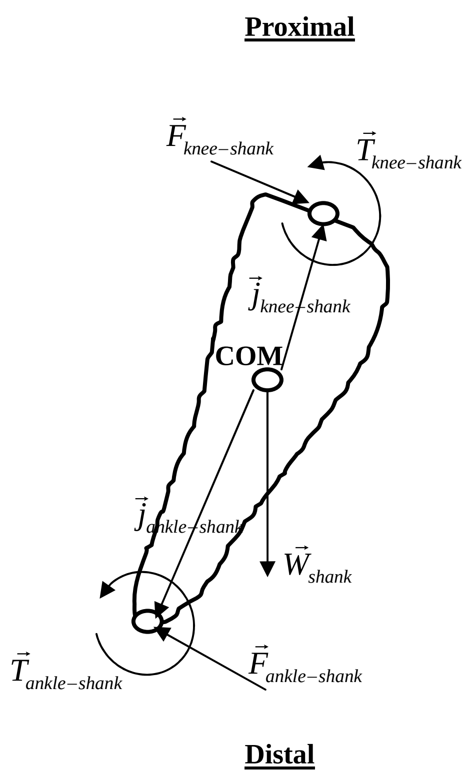

Figure 3.4-4 Free body diagram of shank segment

A free body diagram for right shank segment was illustrated in Figure 3.4-4. All the related quantities were transformed from global coordinate system or local foot coordinate system to local shank coordinate system. There is no environmental force and torque acting on the shank. Therefore expanded equations of motion (3.46)(3.47) were written as

$$\vec{F}_{knee-shank} = m_{shank} \cdot \ddot{\vec{P}}_{shank} - \vec{F}_{ankle-shank} - \vec{W}_{shank} \tag{3.54}$$

$$\vec{T}_{knee-shank} = \dot{\vec{H}}_{shank} - \vec{j}_{ankle-shank} \times \vec{F}_{ankle-shank} - \vec{T}_{ankle-shank} - \vec{j}_{knee-shank} \times \vec{F}_{knee-shank} \tag{3.55}$$

Considering the Newton's third law,

$$\vec{F}_{ankle-shank} = -\vec{F}_{ankle-foot} \tag{3.56}$$

$$\vec{T}_{ankle-shank} = -\vec{T}_{ankle-foot} \tag{3.57}$$

Thus, Eq. (3.54)(3.55) were reduced to

$$\vec{F}_{knee-shank} = m_{shank} \cdot \ddot{\vec{P}}_{shank} + \vec{F}_{ankle-foot} - \vec{W}_{shank} \tag{3.58}$$

$$\vec{T}_{knee-shank} = \dot{\vec{H}}_{shank} + \vec{j}_{ankle-shank} \times \vec{F}_{ankle-foot} + \vec{T}_{ankle-foot} - \vec{j}_{knee-shank} \times \vec{F}_{knee-shank} \tag{3.59}$$

The resultant joint moment $\vec{T}_{knee-shank}$ was reported in local shank coordinate system. In addition, to facilitate further computation of the hip joint moment, this shank joint moment $\vec{T}_{knee-shank}$ was transformed into local thigh coordinate system.

Solution for the right thigh segment

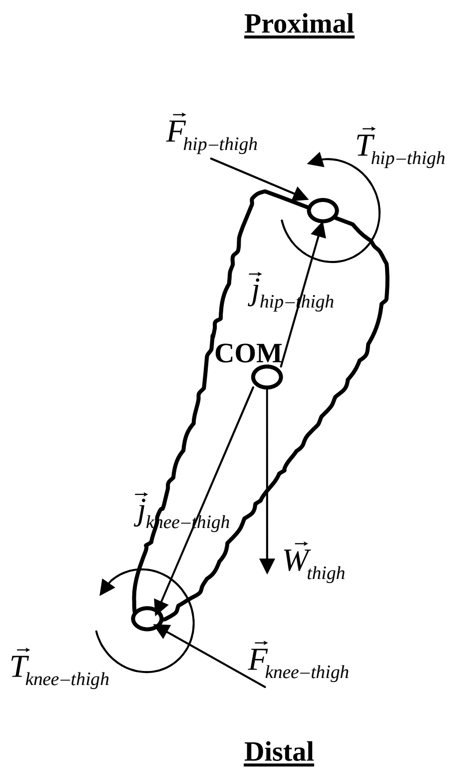

Figure 3.4-5 Free body diagram of thigh segment

A free body diagram for right thigh segment was illustrated in Figure 3.4-5. All the related quantities were transformed into local thigh coordinate system. There is no environmental force and torque acting on the thigh. Therefore expanded equations of motion (3.46)(3.47) were written as

$$\vec{F}_{hip-thigh} = m_{thigh} \cdot \ddot{\vec{P}}_{thigh} + \vec{F}_{knee-shank} - \vec{W}_{thigh} \tag{3.60}$$

$$\vec{T}_{hip-thigh} = \dot{\vec{H}}_{thigh} - \vec{j}_{knee-thigh} \times \vec{F}_{knee-thigh} - \vec{T}_{knee-thigh} - \vec{j}_{hip-thigh} \times \vec{F}_{hip-thigh} \tag{3.61}$$

Considering the Newton's third law,

$$\vec{F}_{knee-thigh} = -\vec{F}_{knee-shank} \tag{3.62}$$

$$\vec{T}_{knee-thigh} = -\vec{T}_{knee-shank} \tag{3.63}$$

Thus, Eq. (3.60)(3.61) were further reduced to

$$\vec{F}_{hip-thigh} = m_{thigh} \cdot \ddot{\vec{P}}_{thigh} - \vec{F}_{knee-thigh} - \vec{W}_{thigh} \tag{3.64}$$

$$\vec{T}_{hip-thigh} = \dot{\vec{H}}_{thigh} + \vec{j}_{knee-thigh} \times \vec{F}_{knee-shank} + \vec{T}_{knee-shank} - \vec{j}_{hip-thigh} \times \vec{F}_{hip-thigh} \tag{3.65}$$

The resultant joint moment $\vec{T}_{hip-thigh}$ was reported in local thigh coordinate system.

3.5. Data analysis

Descriptive and inferential statistical analyses were performed in the JMP and SAS statistical packages (SAS Institute Inc. Cary, NC, USA). A significant level of $p \le 0.05$ will be used throughout the test.

The time-related dependent variables, JMA Time, JMAP Time and JMG Rate, were analyzed using one-way between-subjects ANCOVA with effect of age groups, considering walking velocity as a covariate. The ratio- and magnitude-related dependent variable, JMP Magnitude, JMP Ratio, JMG Ratio and JMP Power, were analyzed using two-way mixed-subjects ANCOVA with age groups as between-subject factor and gait conditions as within-subject factor, considering walking velocity as a covariate. Walking velocity was involved as a covariate to count for in possible influence on joint moment generation (White & Lage, 1993).

Walking velocity and step length in normal walking data sets were analyzed using one-way between-subjects ANOVA with levels of year 2002 and 2003. Bivariate correlation analysis was used to analyze the relationship of three exertion levels ($30^\circ/s$, $60^\circ/s$ and $120^\circ/s$) as for maximum isokinetic joint torque measurements. Dependent variables obtained were preprocessed by normality check and outlier detection. Box-and-whisker plot and residue plot were applied to assist outlier detection.

Chapter 4

4. RESULTS

4.1. Average joint moment profile in normal walking

In normal walking, consistent joint moment patterns in lower extremity joints were observed in all the participants. At the ankle joint, sagittal joint moments generated characterized by an initial small dorsiflexor (~20-25% of stance) followed by a major plantarflexor during most of the stance (Figure 4.1-2). Ankle frontal and transverse joint moments generated were of small magnitude and highly variable (Figure 4.1-1 and Figure 4.1-3). At the knee joint, sagittal joint moment showed flexor and extensor alternatively (Figure 4.1-5). At the hip joint, sagittal joint moment was dominated by extensor in the first half of stance then turned into flexor in the second half of stance (Figure 4.1-8). Frontal joint moments generated at the knee and hip joints were characterized by two abductor peaks (Figure 4.1-4 and Figure 4.1-7), similar to the normal gait ground reaction profile. Ensemble averages of joint moment curves across all the 18 participants in normal walking were plotted below at each location (reference plane by joint).

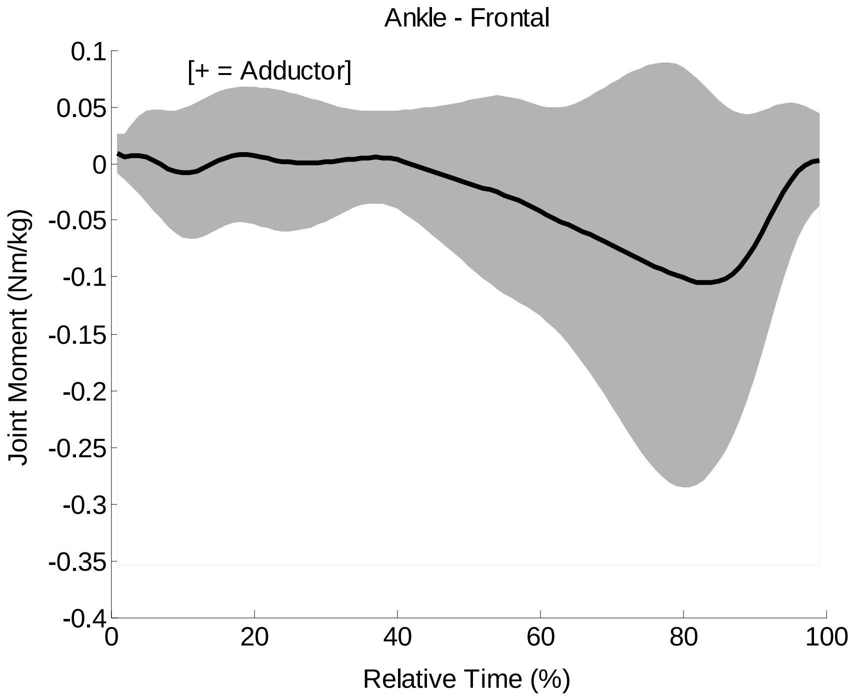

Figure 4.1-1 Stance phase joint moment profile in normal walking (shaded area represent +/- 1 SD)

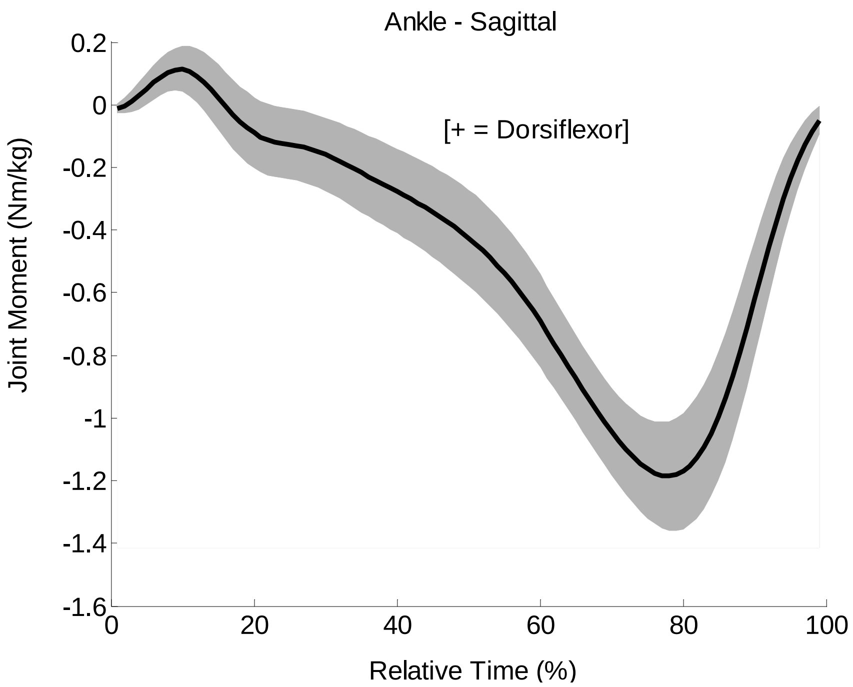

Figure 4.1-2 Stance phase joint moment profile in normal walking (shaded area represent +/- 1 SD)

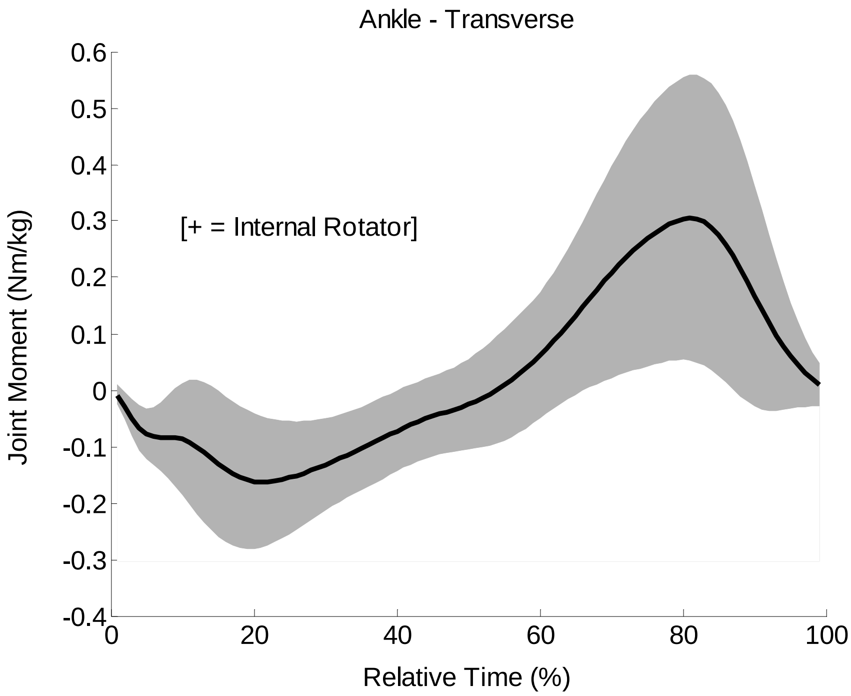

Figure 4.1-3 Stance phase joint moment profile in normal walking (shaded area represent +/- 1 SD)

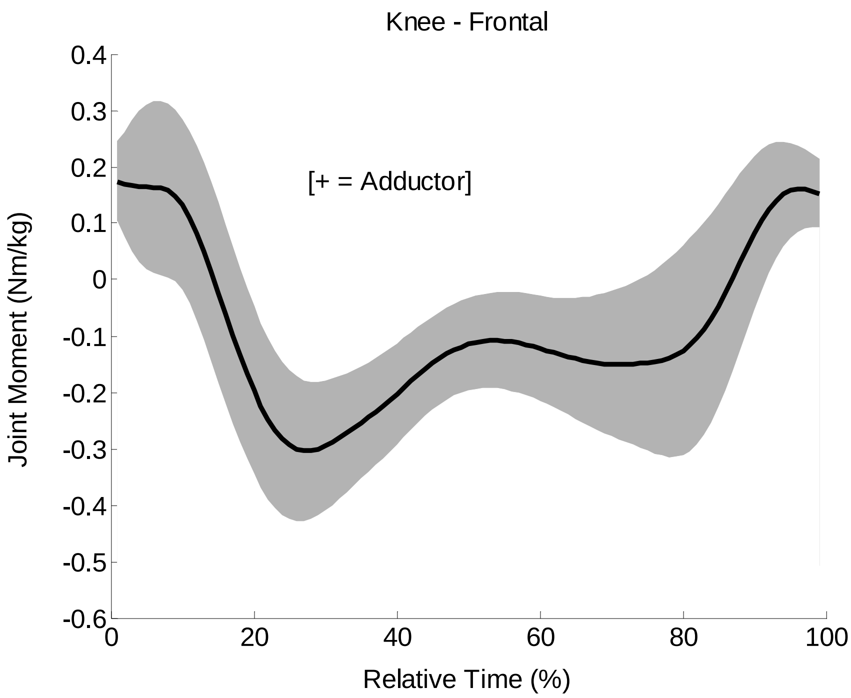

Figure 4.1-4 Stance phase joint moment profile in normal walking (shaded area represent +/- 1 SD)

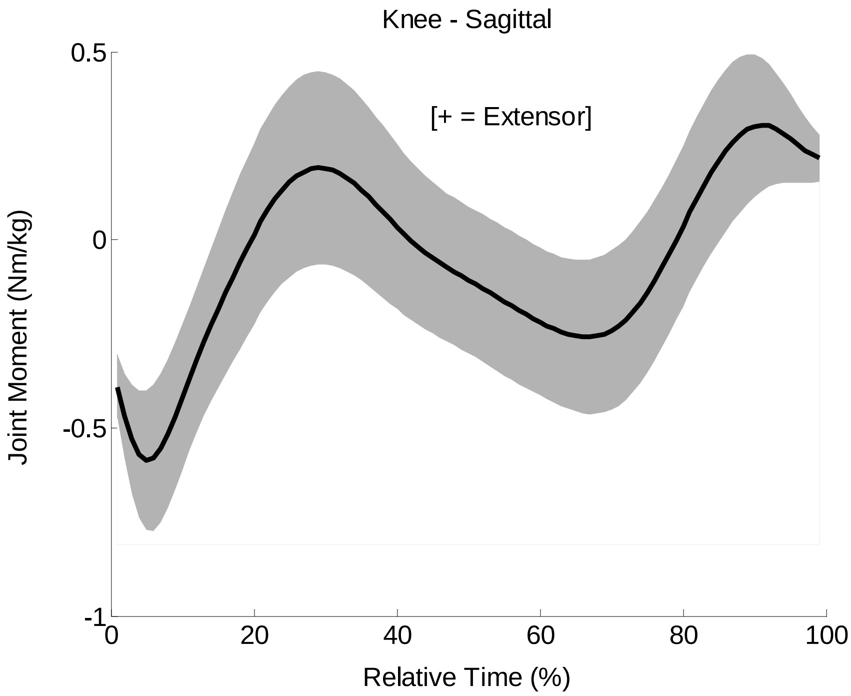

Figure 4.1-5 Stance phase joint moment profile in normal walking (shaded area represent +/- 1 SD)

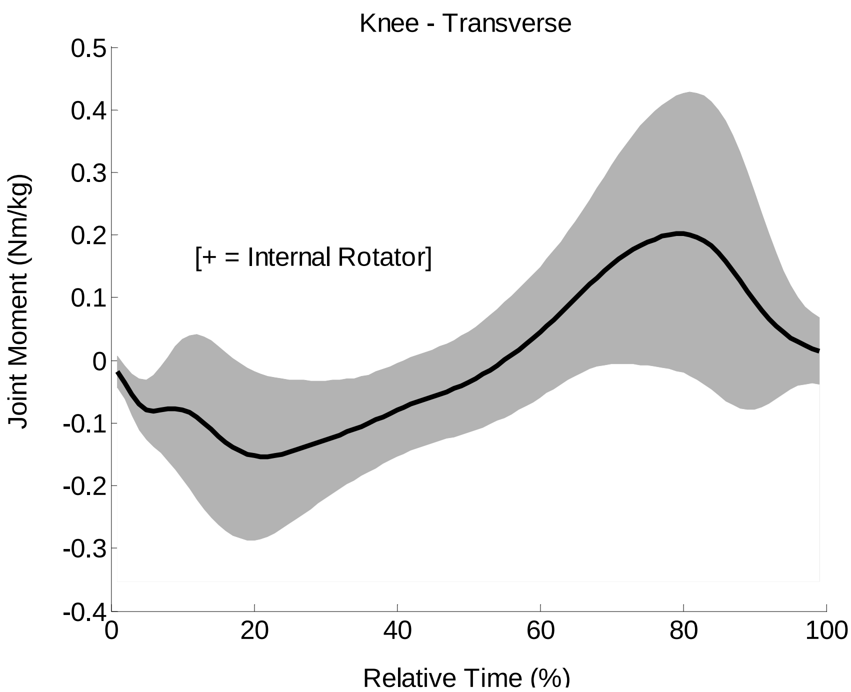

Figure 4.1-6 Stance phase joint moment profile in normal walking (shaded area represent +/- 1 SD)

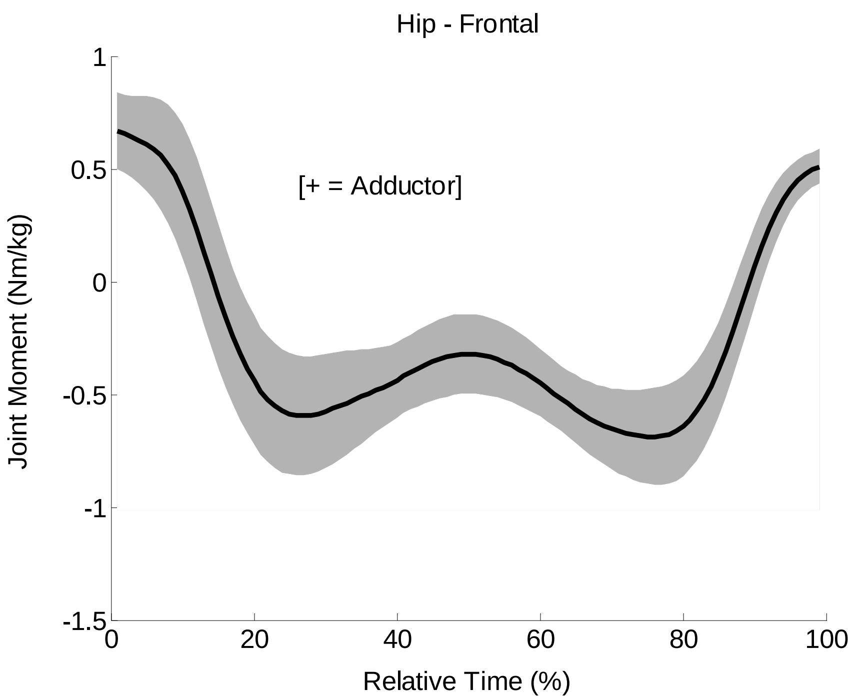

Figure 4.1-7 Stance phase joint moment profile in normal walking (shaded area represent +/- 1 SD)

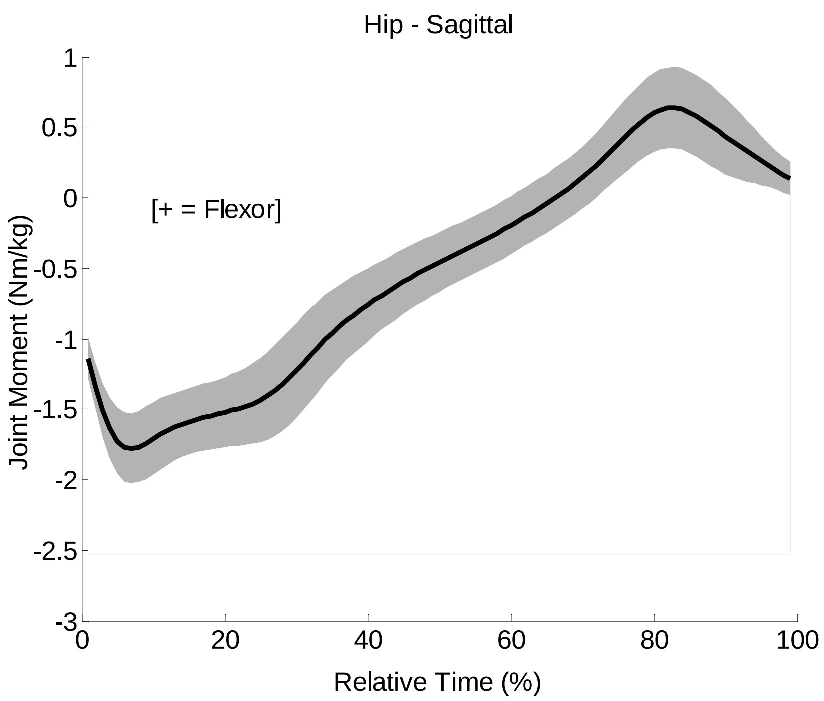

Figure 4.1-8 Stance phase joint moment profile in normal walking (shaded area represent +/- 1 SD)

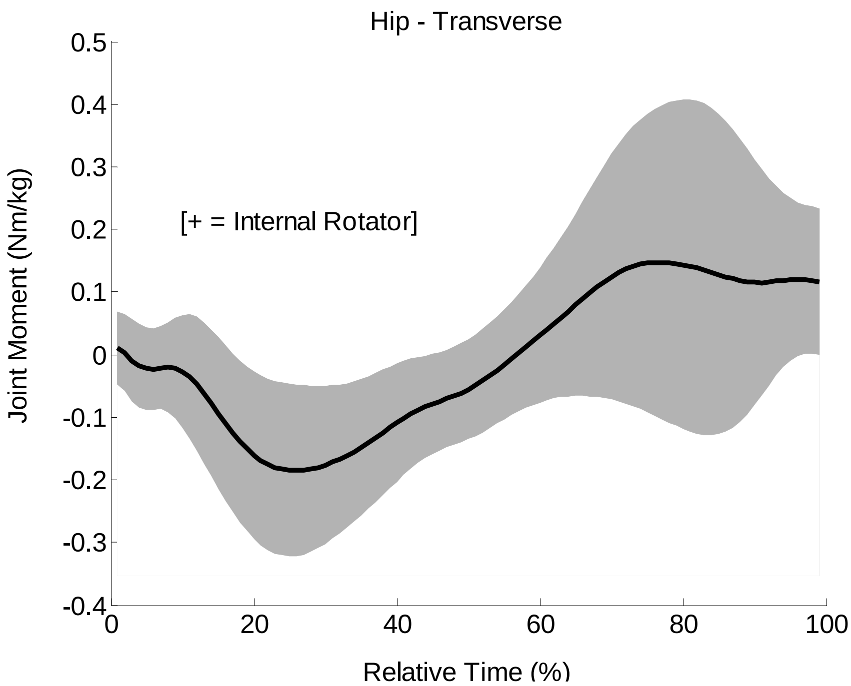

Figure 4.1-9 Stance phase joint moment profile in normal walking (shaded area represent +/- 1 SD)

4.2. Typical joint moment profile in successful reactive-recovery

During successful reactive-recovery, joint moments generated were significantly different than those computed during normal walking. At the ankle joint, sagittal joint moments generated after initial silent phase (~20%-25% of stance) were dominated by dorsiflexor (Figure 4.2-1). At the knee joint, sagittal joint moments generated after initial silent phase were governed by extensor (Figure 4.2-2). At the hip joint, sagittal joint moments generated characterized by two major extensor phases, separated by a temporarily gap (Figure 4.2-3). In the frontal and transverse planes of ankle, knee and hip joints, joint moments generated after initial silent phase rapidly reach peak joint moment; though the dominant muscle groups varied corresponding to individual-specific reactive-recovery process. Because of the highly variable dominant muscle groups in the frontal and transverse plane, only one typical joint moment profile is presented below at each joint.

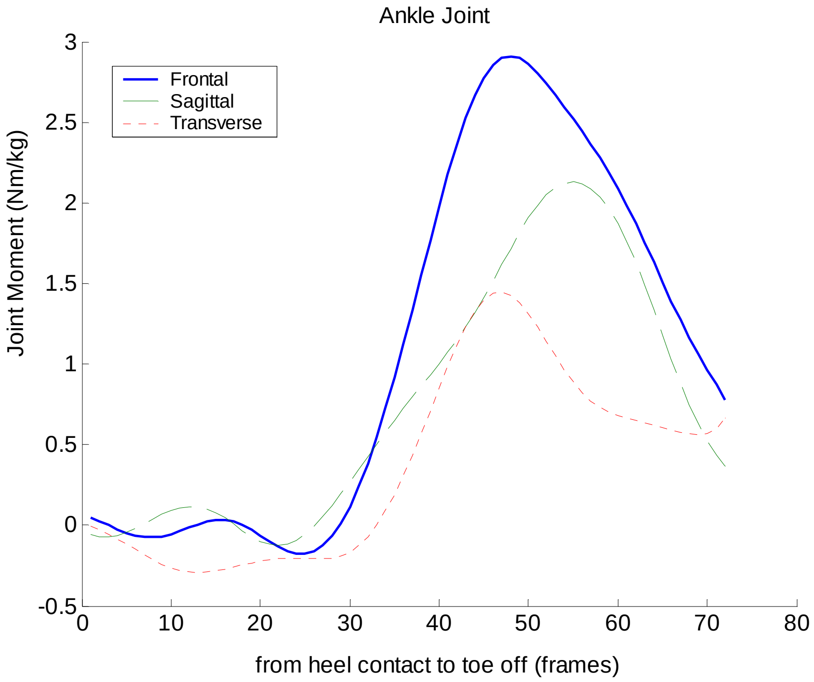

Figure 4.2-1 Typical joint moment profile during reactive-recovery ("+" frontal = invertor; "+" sagittal = dorsiflexor; "+" transverse = internal rotator); 1 frame = 1/120 seconds.

Figure 4.2-2 Typical joint moment profile during reactive-recovery ("+" frontal = adductor; "+" sagittal =

extensor; "+" transverse = internal rotator); 1 frame = 1/120 seconds.

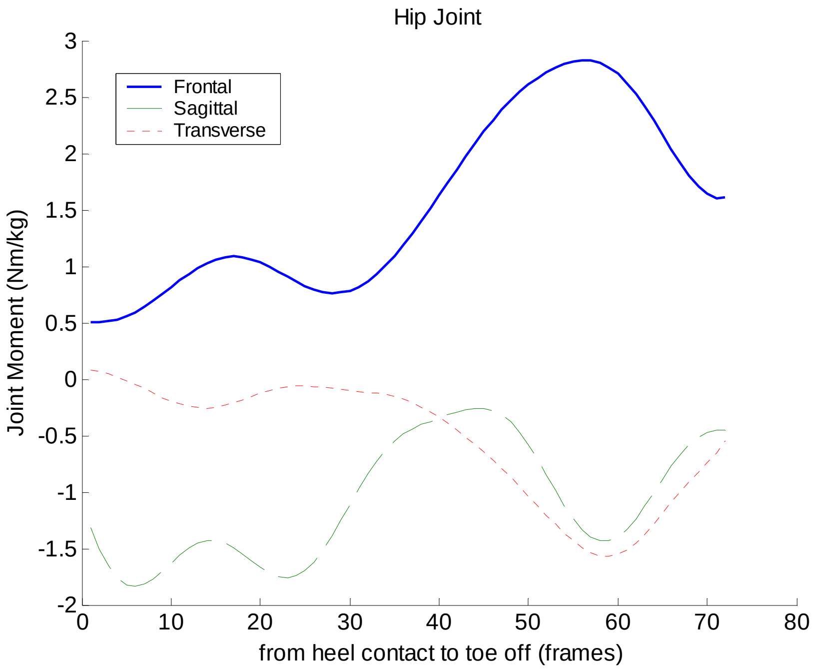

Figure 4.2-3 Typical joint moment profile during reactive-recovery ("+" frontal = adductor; "+" sagittal = flexor; "+" transverse = internal rotator); 1 frame = 1/120 seconds.

4.3. JMA Time (joint moment activation time)

JMA Time in all the locations was analyzed using one-way between-subjects ANCOVA with two levels: young and old. Walking velocity was considered as a covariate. Descriptive summary of JMA Time was shown in Table 4.3-1. Hip sagittal plane location was excluded from the analysis because JMA Time concept was incompatible with the unique double-peak joint moment profile detected in hip sagittal plane.

At the ankle joint, there was no significant effect of age group in the frontal plane ($F(1,13) = 2.28$, $p = 0.155$), in the sagittal plane ($F(1,15) = 0.50$, $p = 0.492$), or in the transverse plane ($F(1,15) = 2.01$, $p = 0.177$).

At the knee joint, there was no significant effect of age group in the frontal plane (F(1,14) = 1.47, p = 0.245), in the sagittal plane (F(1,10) = 0.09, p = 0.771), or in the transverse plane (F(1,15) = 2.37, p = 0.144).

Table 4.3-1 Means and SD of JMA Time (ms)

Joint	Reference Plane	Old $\bar{X}(SD)$	Young $\bar{X}(SD)$	Effect
Ankle	Frontal	259.26 (56.74)	269.05 (20.81)	None
	Sagittal	273.15 (99.11)	275.00 (58.93)	None
	Transverse	257.41 (63.80)	303.70 (87.51)	None
Knee	Frontal	238.89 (65.88)	261.46 (40.08)	None
	Sagittal	289.58 (24.70)	276.67 (3.73)	None
	Transverse	231.25 (43.13)	293.52 (87.08)	None
Hip	Frontal	239.82 (74.63)	276.85 (80.56)	None
	Transverse	293.75 (88.05)	297.22 (66.14)	None

At the hip joint, there was no significant effect of age group in the frontal plane (F(1,15) = 3.25, p = 0.092), or in the transverse plane (F(1,14) = 0.20, p = 0.665).

Therefore, during the successful reactive-recovery, both young and old individuals demanded the same amount of time to initiate reactive joint moments.

4.4. JMAP Time (joint moment activation to peak time)

JMAP Time in all the locations was analyzed using one-way between-subjects ANCOVA with two levels: young and old. Walking velocity was considered as a covariate. Descriptive summary of JMAP Time was shown in Table 4.4-1. Hip sagittal plane location was excluded from the analysis because JMAP Time concept was incompatible with the unique double-peak joint moment profile detected in hip sagittal plane.

At the ankle joint, there was no significant effect of age group in the frontal plane (F(1,15) = 3.78, p = 0.0708), in the sagittal plane (F(1,15) = 0.20, p = 0.660), or in the transverse plane (F(1,15) = 1.06, p = 0.320).

At the knee joint, there was no significant effect of age group in the frontal plane ($F(1,15) = 0.60$, $p = 0.450$), in the sagittal plane ($F(1,15) < 0.01$, $p = 0.949$), or in the transverse plane ($F(1,15) = 1.05$, $p = 0.321$).

Table 4.4-1 Means and SD of JMAP Time (ms)

Joint	Reference Plane	Old $\bar{X}(SD)$	Young $\bar{X}(SD)$	Effect
Ankle	Frontal	468.52 (74.28)	384.26 (50.94)	None
	Sagittal	464.82 (84.35)	440.74 (61.30)	None
	Transverse	459.26 (93.86)	396.30 (47.89)	None
Knee	Frontal	466.67 (77.73)	412.04 (85.00)	None
	Sagittal	461.11 (59.66)	437.04 (53.54)	None
	Transverse	476.85 (100.67)	400.93 (60.87)	None
Hip	Frontal	427.78 (90.72)	399.08 (71.98)	None
	Transverse	487.96 (122.34)	462.04 (72.06)	None

At the hip joint, there was no significant effect of age group in the frontal plane ($F(1,15) = 0.02$, $p = 0.889$), or in the transverse plane ($F(1,15) = 0.14$, $p = 0.711$).

Therefore, during successful reactive-recovery, both young and old individuals required the same amount of time to reach peak joint moment, after the reactive joint moment was initiated.

4.5. JMP Magnitude (peak joint moment magnitude)

JMP Magnitude in all locations (joint by reference plane) was analyzed using two-way mixed-subjects ANCOVA with four conditions: old/normal, old/recovery, young/normal, young/recovery. Walking velocity was considered as a covariate.

4.5.1. Ankle frontal plane

Table 4.5-1 Means and SD of ankle frontal JMP Magnitude (Nm/kg)

	Old $\bar{X}(SD)$	Young $\bar{X}(SD)$
normal	0.23 (0.06)	0.27 (0.10)
reactive-recovery	1.62 (0.64)	1.24 (0.84)

There was a significant main effect of gait condition ($F(1,15) = 36.00$, $p < 0.001$) and no significant main effect of age group ($F(1,14.8) = 0.80$, $p = 0.379$). The interaction was not significant ($F(1,15) = 1.24$, $p = 0.284$).

Thus, both young and old groups generated greater peak joint moments in ankle frontal plane during successful reactive-recovery than during normal walking. This is illustrated in Figure 4.5-1.

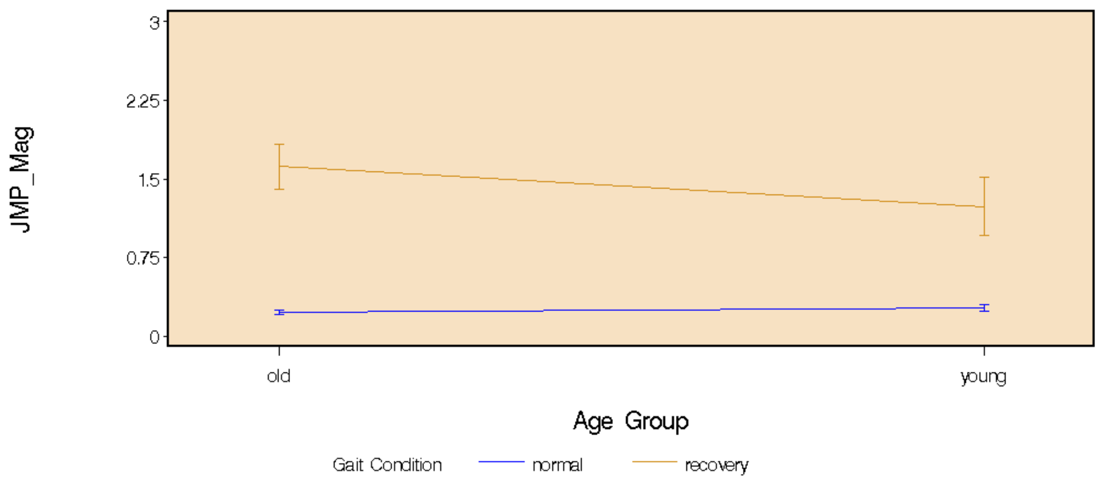

Figure 4.5-1 Interaction effect on ankle frontal JMP Magnitude

4.5.2. *Ankle sagittal plane*

Table 4.5-2 Means and SD of ankle sagittal JMP Magnitude (Nm/kg)

	Old $\bar{X}\,(SD)$	Young $\bar{X}\,(SD)$
normal	1.35 (0.09)	1.51 (0.17)
reactive-recovery	1.92 (1.33)	3.66 (1.76)

There was a significant main effect of gait condition ($F(1,15.5) = 12.84$, $p = 0.003$) and a significant main effect of age group ($F(1,15.5) = 6.10$, $p = 0.026$). The interaction was not significant ($F(1,15.5) = 4.40$, $p = 0.053$).

Hence, both age groups were found to develop greater peak joint moment during reactive-recovery than during normal walking. Meanwhile, in both gait conditions, the younger group was found to produce greater peak joint moment than their older counterparts. This is illustrated in Figure 4.5-2.

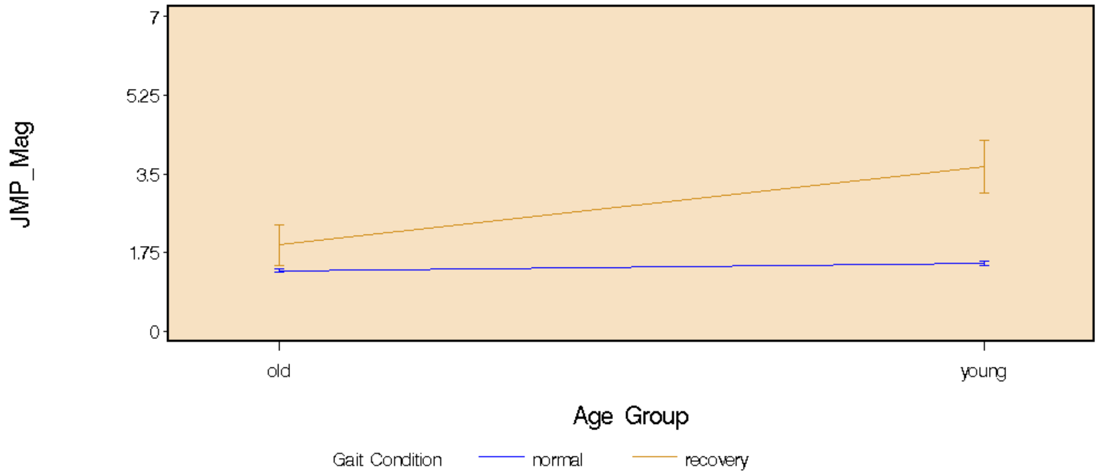

Figure 4.5-2 Interaction effect on ankle sagittal JMP Magnitude

4.5.3. Ankle transverse plane

Table 4.5-3 Means and SD of ankle transverse JMP Magnitude (Nm/kg)

	Old $\bar{X}$ (*SD*)	Young $\bar{X}$ (*SD*)
normal	0.27 (0.06)	0.30 (0.10)
reactive-recovery	0.95 (0.58)	0.67 (0.44)

There was a significant main effect of gait condition ($F(1,10.4) = 15.38$, $p = 0.003$) and no significant main effect of age group ($F(1,10.2) = 0.85$, $p = 0.378$). The interaction was not significant ($F(1,10.4) = 1.35$, $p = 0.271$).

Thus, both young and old groups generated greater peak joint moment in ankle transverse plane during successful reactive-recovery, with respect to normal walking. This is illustrated in Figure 4.5-3.

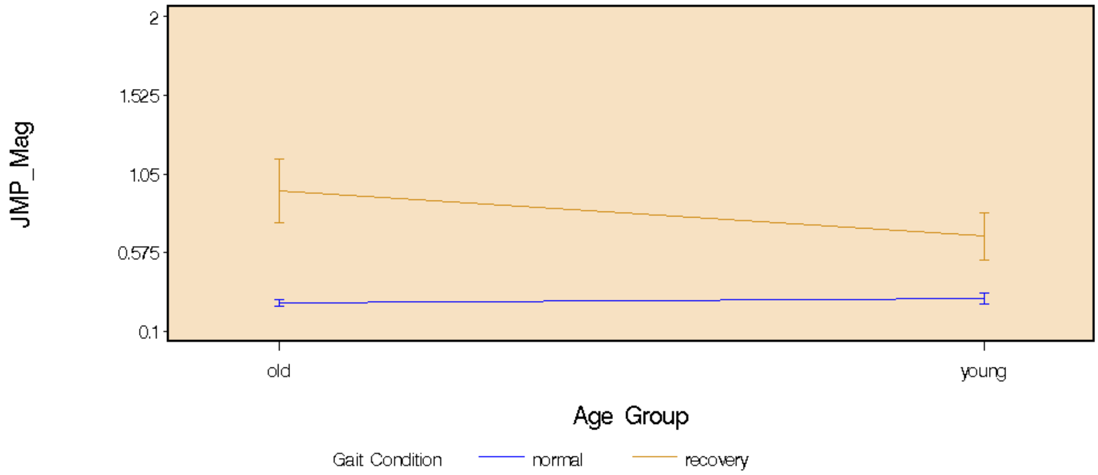

Figure 4.5-3 Interaction effect on ankle transverse JMP Magnitude

4.5.4. Knee frontal plane

Table 4.5-4 Means and SD of knee frontal JMP Magnitude (Nm/kg)

	Old $\bar{X}$ (*SD*)	Young $\bar{X}$ (*SD*)
normal	0.53 (0.13)	0.59 (0.08)
reactive-recovery	1.52 (0.63)	1.26 (0.37)

There was a significant main effect of gait condition ($F(1,16.2) = 41.93$, $p < 0.001$) and no significant main effect of age group ($F(1,16.2) = 0.60$, $p = 0.448$). The interaction was not significant ($F(1,16.2) = 1.50$, $p = 0.239$).

Thus, both young and old groups generated greater peak joint moment in knee frontal plane during successful reactive-recovery, with respect to normal walking. This is illustrated in Figure 4.5-4.

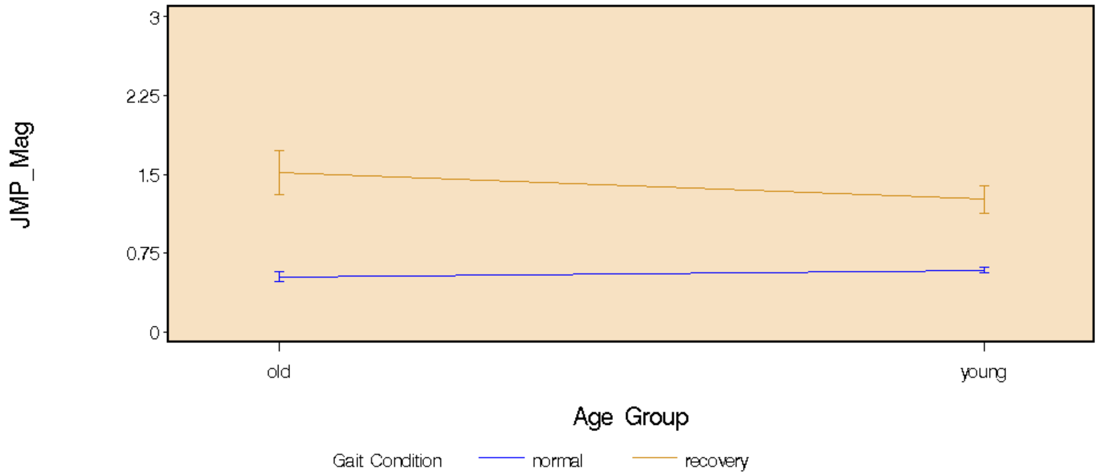

Figure 4.5-4 Interaction effect on knee frontal JMP Magnitude

4.5.5. Knee sagittal plane

Table 4.5-5 Means and SD of knee sagittal JMP Magnitude (Nm/kg)

	Old $\bar{X}\,(SD)$	Young $\bar{X}\,(SD)$
normal	0.61 (0.09)	0.63 (0.09)
reactive-recovery	1.39 (0.64)	3.33 (1.72)

There was a significant main effect of gait condition (F(1,6.35) = 25.02, p = 0.002) and a significant main effect of age group (F(1,6.06) = 10.42, p = 0.018). However, the interaction was also significant (F(1,6.35) = 7.41, p = 0.033).

Looking at the simple main effects, there was a significant effect of gait condition for both young (F(1,15) = 11.66, p = 0.004) and old groups (F(1,15) = 19.38, p = 0.001). There was a significant effect of age group during reactive-recovery (F(1,16) = 9.94, p = 0.006), but not during normal walking (F(1,14) = 0.31, p = 0.589).

Hence, both young and old participants were found to generate greater peak joint moments in knee sagittal plane during successful reactive-recovery, with respect to normal walking. However, only during reactive-recovery, younger group produced more peak joint moments than the elderly group. This is illustrated in Figure 4.5-5.

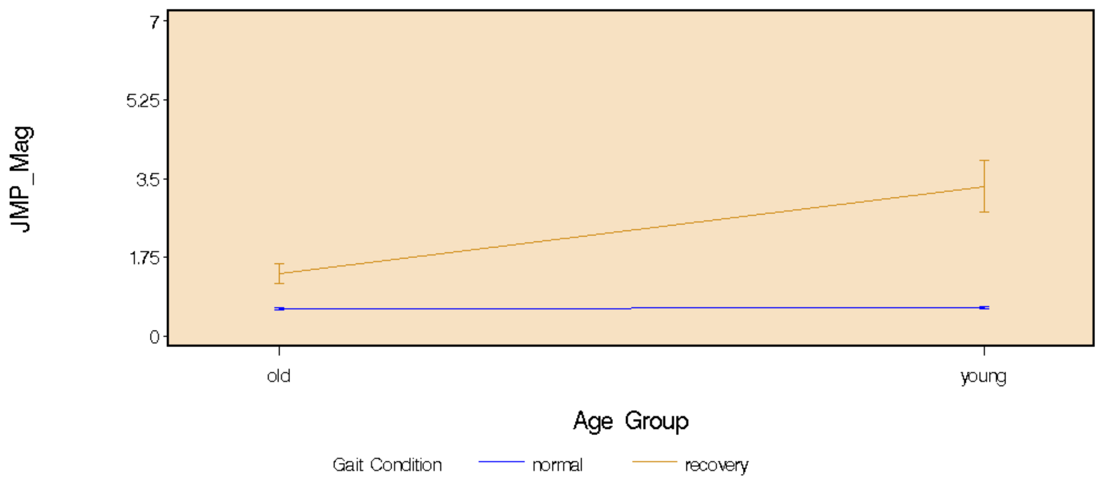

Figure 4.5-5 Interaction effect on knee sagittal JMP Magnitude

4.5.6. Knee transverse plane

Table 4.5-6 Means and SD of knee transverse JMP Magnitude (Nm/kg)

	Old $\bar{X}\,(SD)$	Young $\bar{X}\,(SD)$
normal	0.25 (0.13)	0.20 (0.08)
reactive-recovery	0.74 (0.25)	0.36 (0.23)

There was a significant main effect of gait condition (F(1,14.8) = 17.34, p < 0.001) and a significant main effect of age group (F(1,12.9) = 14.48, p = 0.002). However, the interaction was also significant (F(1,14.8) = 5.32, p = 0.036).

Looking at the simple main effects, there was a significant effect of gait condition for the elderly group (F(1,14) = 25.30, p < 0.001), but not for the young group (F(1,14) = 3.61, p = 0.078). There was a significant effect of age group during reactive-recovery (F(1,14) = 10.41, p = 0.006), but not during normal walking (F(1,14) = 0.78, p = 0.393).

Therefore, only the elderly group was found to generate greater peak joint moment during successful reactive-recovery, with respect to normal walking. Meanwhile, only during reactive-recovery, the elderly group was found to produce more peak joint moment than their younger counterparts. This is illustrated in Figure 4.5-6.

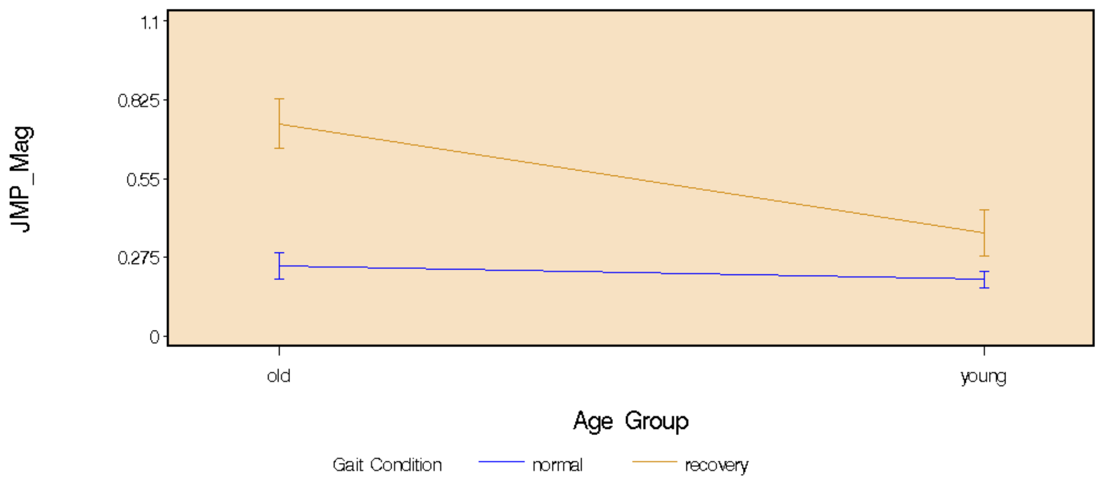

Figure 4.5-6 Interaction effect on knee transverse JMP Magnitude

4.5.7. Hip frontal plane

Table 4.5-7 Means and SD of hip frontal JMP Magnitude (Nm/kg)

	Old $\bar{X}\,(SD)$	Young $\bar{X}\,(SD)$
normal	1.80 (0.14)	1.81 (0.04)
reactive-recovery	2.54 (0.26)	2.23 (0.52)

There was a significant main effect of gait condition ($F(1,8.25) = 32.74$, $p < 0.001$) and no significant main effect of age group ($F(1,8.24) = 2.04$, $p = 0.190$). The interaction was not significant ($F(1,8.24) = 2.57$, $p = 0.147$).

Thus, both young and old groups generated greater peak joint moment in hip frontal plane during successful reactive-recovery, with respect to normal walking. This is illustrated in Figure 4.5-7.

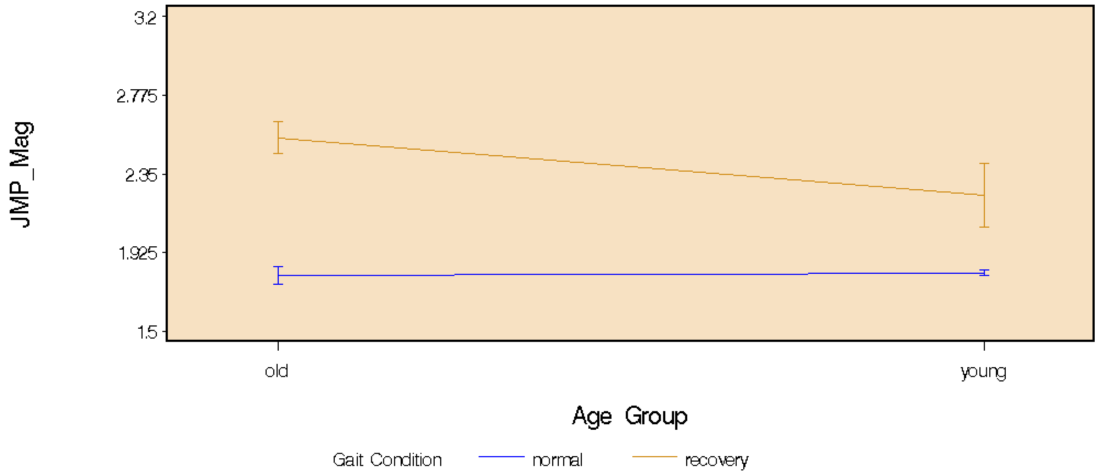

Figure 4.5-7 Interaction effect on hip frontal JMP Magnitude

4.5.8. *Hip sagittal plane*

Table 4.5-8 Means and SD of hip sagittal JMP Magnitude (Nm/kg)

	Old $\bar{X}(SD)$	Young $\bar{X}(SD)$
normal	1.88 (0.17)	1.86 (0.19)
reactive-recovery	1.26 (0.37)	1.28 (0.17)

There was a significant main effect of gait condition (F(1,16.4) = 41.13, p < 0.001) and no significant main effect of age group (F(1,16) < 0.01, p = 0.993). The interaction was not significant (F(1,16.4) = 0.070, p = 0.798).

Thus, both young and old groups generated greater peak joint moment in hip sagittal plane during normal walking, with respect to successful reactive-recovery. This is illustrated in Figure 4.5-8.

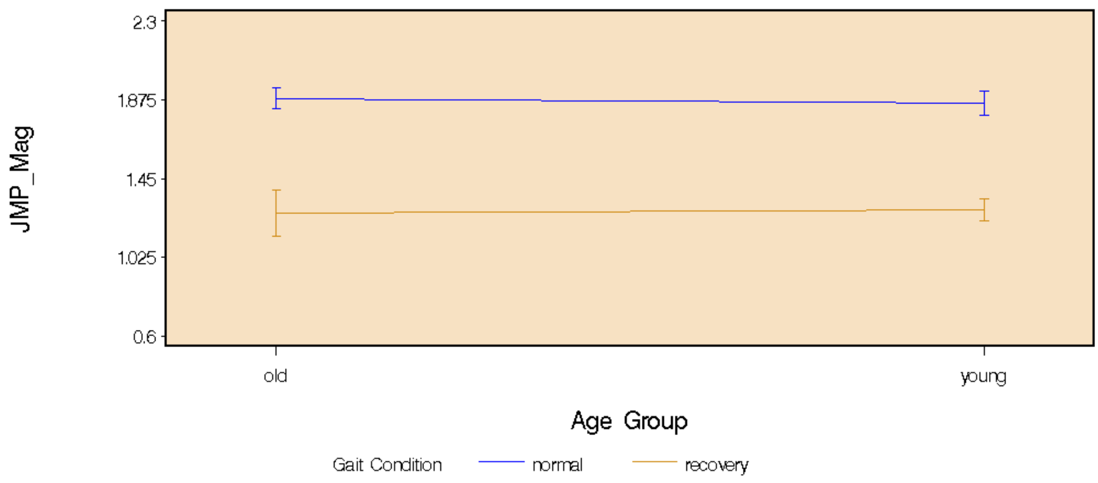

Figure 4.5-8 Interaction effect on hip sagittal JMP Magnitude

4.5.9. Hip transverse plane

Table 4.5-9 Means and SD of hip transverse JMP Magnitude (Nm/kg)

	Old $\bar{X}\,(SD)$	Young $\bar{X}\,(SD)$
normal	0.39 (0.13)	0.32 (0.09)
reactive-recovery	0.50 (0.30)	0.91 (0.38)

There was a significant main effect of gait condition ($F(1,16) = 17.36$, $p < 0.001$) and no significant main effect of age group ($F(1,16.1) = 3.62$, $p = 0.075$). However, the interaction was significant ($F(1,16) = 7.76$, $p = 0.013$).

Looking at the simple main effects, there was a significant effect of gait condition for younger group ($F(1,14) = 18.61$, $p < 0.001$), but not elderly group ($F(1,16) = 1.18$, $p = 0.293$), showing that younger people generated greater peak joint moments in both gait conditions.

Therefore, only younger people generated greater peak joint moment in hip transverse plane during successful reactive-recovery, with respect to normal walking. This is illustrated in Figure 4.5-9.

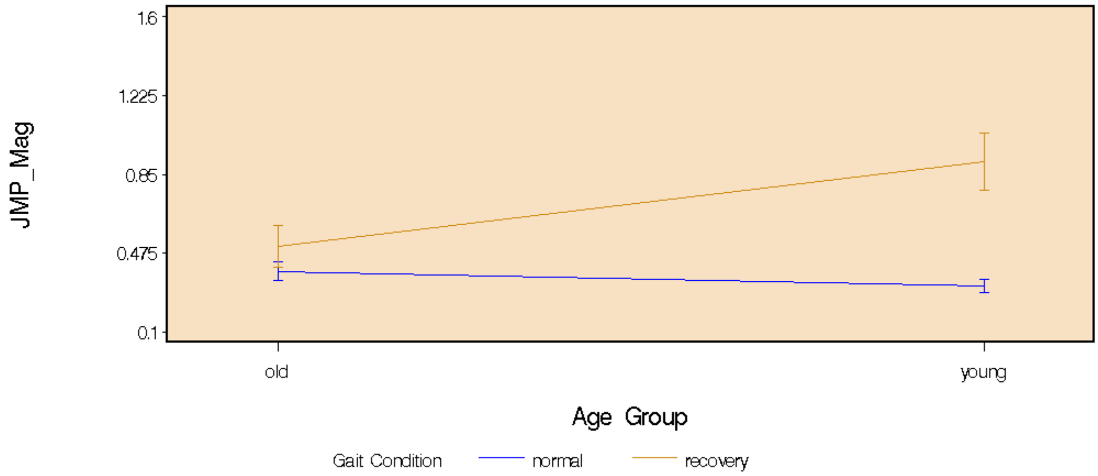

Figure 4.5-9 Interaction effect on hip transverse JMP Magnitude

4.6. JMG Ratio (joint moment generation ratio)

JMG Ratio in ankle sagittal and knee sagittal planes was analyzed using two-way mixed-subjects ANCOVA with four conditions: old/normal, old/recovery, young/normal, and young/recovery. Walking velocity was considered as a covariate.

4.6.1. Ankle sagittal plane

Table 4.6-1 Means and SD of ankle sagittal JMG Ratio

	Old $\bar{X}\,(SD)$	Young $\bar{X}\,(SD)$
normal	3.08 (0.93)	3.37 (0.99)
recovery	6.34 (5.15)	8.36 (4.78)

There was a significant main effect of gait condition ($F(1,6.81) = 9.61$, $p = 0.018$) and no significant main effect of age group ($F(1,7.19) = 0.23$, $p = 0.645$). The interaction was not significant ($F(1,6.81) = 0.83$, $p = 0.394$).

Hence, both age groups were found to increase their peak joint moments relative to their strength measurements in ankle sagittal plane during successful reactive-recovery condition, with respect to normal walking. This is illustrated in Figure 4.6-1.

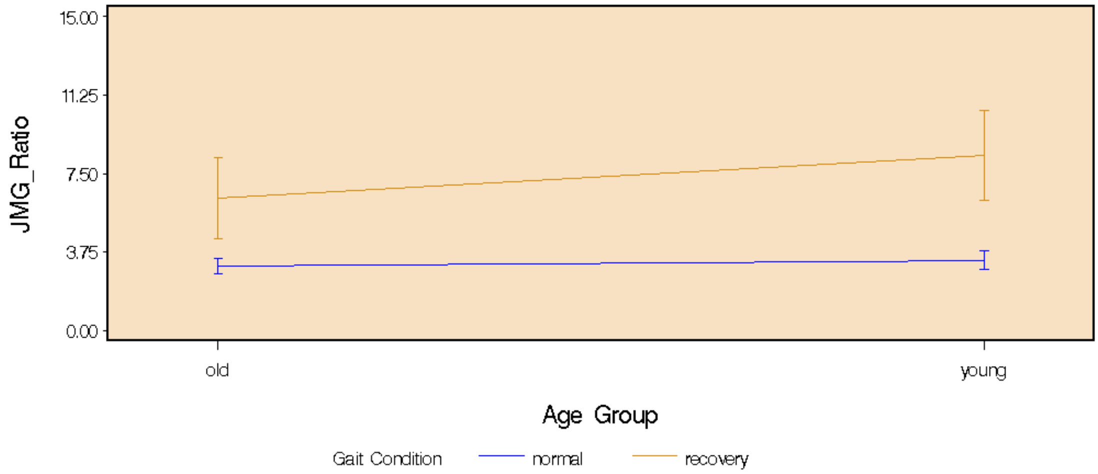

Figure 4.6-1 Interaction effect on ankle sagittal JMG Ratio

4.6.2. Knee sagittal plane

Table 4.6-2 Means and SD of knee sagittal JMG Ratio

	Old $\bar{X}\,(SD)$	Young $\bar{X}\,(SD)$
normal	0.91 (0.50)	0.63 (0.17)
recovery	1.77 (1.18)	3.21 (1.36)

There was a significant main effect of gait condition (F(1,10) = 26.69, p < 0.001) and no significant main effect of age group (F(1,10) = 1.78, p = 0.212). However, the interaction was also significant (F(1,10) = 6.70, p = 0.027).

Looking at the simple main effects, there was a significant effect of gait condition for the younger group (F(1,8) = 17.69, p = 0.003), but not for the elderly group (F(1,12) = 3.14, p = 0.102) showing that only the younger group produced higher JMG Ratio during reactive-recovery than during normal walking.

Thus, only young participants were found to generate greater ratio of peak joint moments to their strength measurements in knee sagittal plane, during successful reactive-recovery than during normal walking. This is illustrated in Figure 4.6-2.

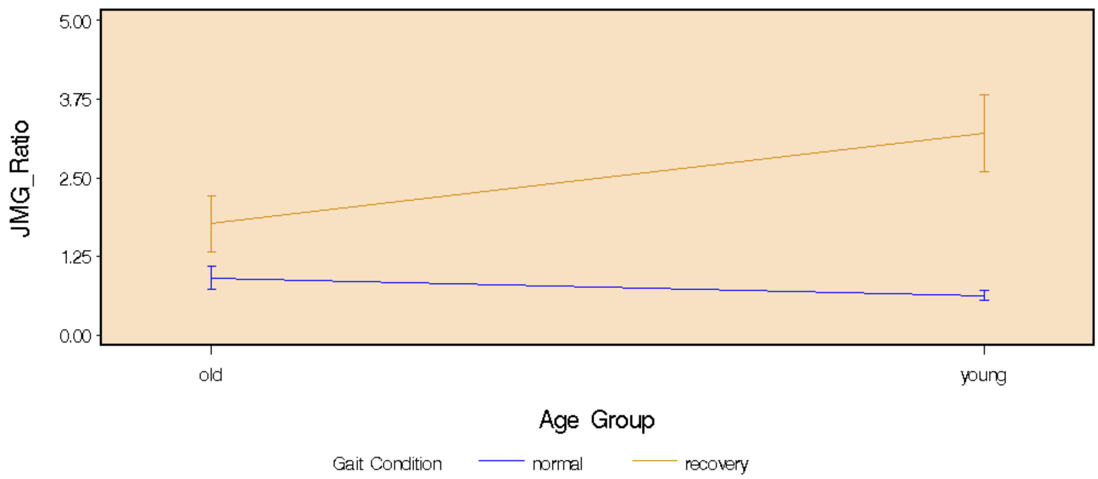

Figure 4.6-2 Interaction effect on knee sagittal JMG Ratio

4.7. JMP Ratio (peak joint moment ratio)

JMP Ratio in all locations (joint*reference plane) was analyzed using two-way mixed-subjects ANCOVA with four conditions: old/normal, old/recovery, young/normal, young/recovery. Walking velocity was considered as a covariate.

4.7.1. Ankle frontal plane

Table 4.7-1 Means and SD of ankle frontal JMP Ratio (Percent)

	Old $\bar{X}$ (SD)	Young $\bar{X}$ (SD)
normal	12.00 (2.65)	14.97 (6.93)
reactive-recovery	41.16 (16.91)	22.66 (13.13)

There was a significant main effect of gait condition ($F(1,16) = 20.76$, $p < 0.001$) and no significant main effect of age group ($F(1,15.9) = 4.35$, $p = 0.054$). However, the interaction was also significant ($F(1,16) = 7.03$, $p = 0.017$).

Looking at the simple main effects, there was a significant effect of gait condition for the old group ($F(1,16) = 23.12$, $p < 0.001$), but not for the younger group ($F(1,16) = 2.42$, $p = 0.140$) showing that the old group possessed more JMP Ratio during reactive-recovery than during normal walking.

Thus, only the elderly was found to increase ankle frontal peak joint moment percentage during successful reactive-recovery, with respect to normal walking. This is illustrated in Figure 4.7-1.

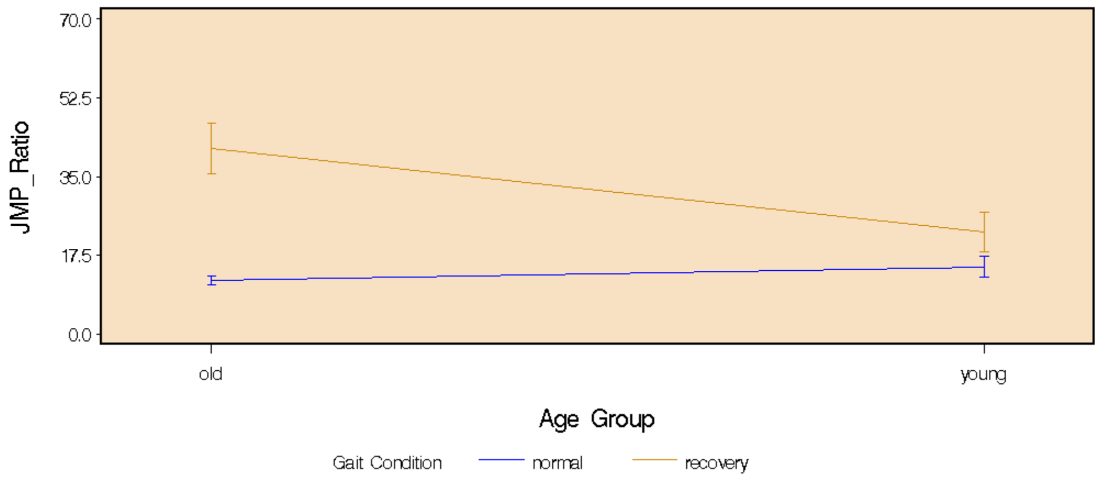

Figure 4.7-1 Interaction effect on ankle frontal JMP Ratio

4.7.2. Ankle sagittal plane

Table 4.7-2 Means and SD of ankle sagittal JMP Ratio (Percent)

	Old $\bar{X}$ (*SD*)	Young $\bar{X}$ (*SD*)
normal	70.67 (6.21)	73.79 (2.85)
reactive-recovery	38.47 (14.24)	64.95 (16.94)

There was a significant main effect of gait condition ($F(1,14.9) = 25.89$, $p < 0.001$) and a significant main effect of age group ($F(1,14.9) = 14.13$, $p = 0.002$). However, the interaction was also significant ($F(1,14.9) = 8.45$, $p = 0.011$).

Looking at the simple main effects, there was a significant effect of age group during reactive-recovery ($F(1,16) = 12.89$, $p = 0.003$), but not during normal walking ($F(1,15) = 1.69$, $p = 0.213$) showing that younger people possessed greater JMP Ratio than the elderly only during reactive-recovery condition. There was a significant effect of gait condition for the elderly group ($F(1,16) = 38.64$, $p < 0.001$), but not for the younger group ($F(1,15) = 2.11$, $p = 0.167$).

Therefore, younger people were found to have more ankle sagittal peak joint moment percentage than their old counterpart only during successful reactive-recovery. Meanwhile, only the elderly reduced their peak joint moment percentage in ankle sagittal plane during recovery condition, with respect to normal walking. This is illustrated in Figure 4.7-2.

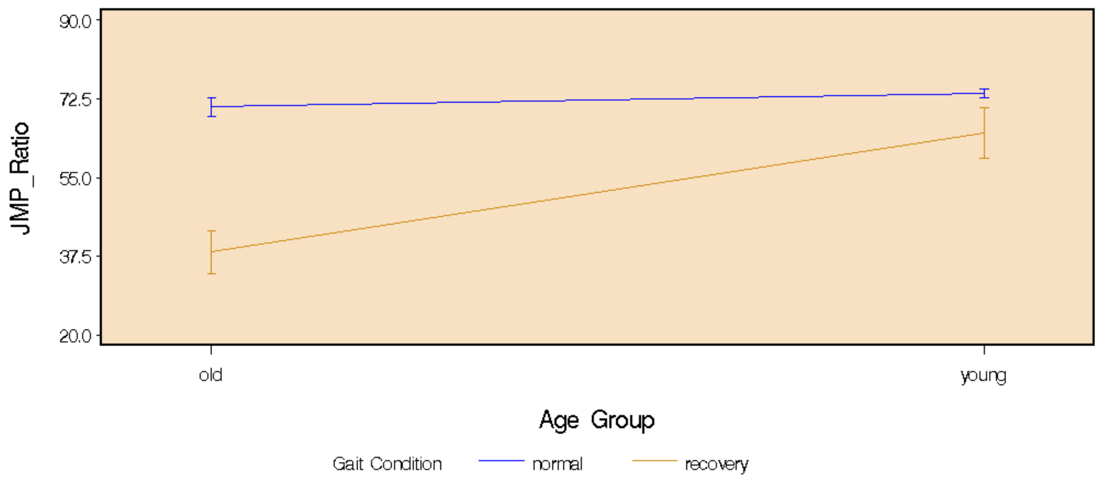

Figure 4.7-2 Interaction effect on ankle sagittal JMP Ratio

4.7.3. *Ankle transverse plane*

Table 4.7-3 Means and SD of ankle transverse JMP Ratio (Percent)

	Old $\bar{X}\,(SD)$	Young $\bar{X}\,(SD)$
normal	16.14 (3.82)	13.81 (3.49)
reactive-recovery	20.37 (5.23)	12.38 (5.81)

There was a significant main effect of age group ($F(1,16) = 9.93$, $p = 0.006$) and no significant main effect of gait condition ($F(1,16) = 0.89$, $p = 0.360$). The interaction was not significant ($F(1,16) = 3.62$, $p = 0.075$).

Thus, the elderly group was found to have more ankle transverse peak joint moment percentage than their younger counterparts, during both gait conditions. This is illustrated in Figure 4.7-3.

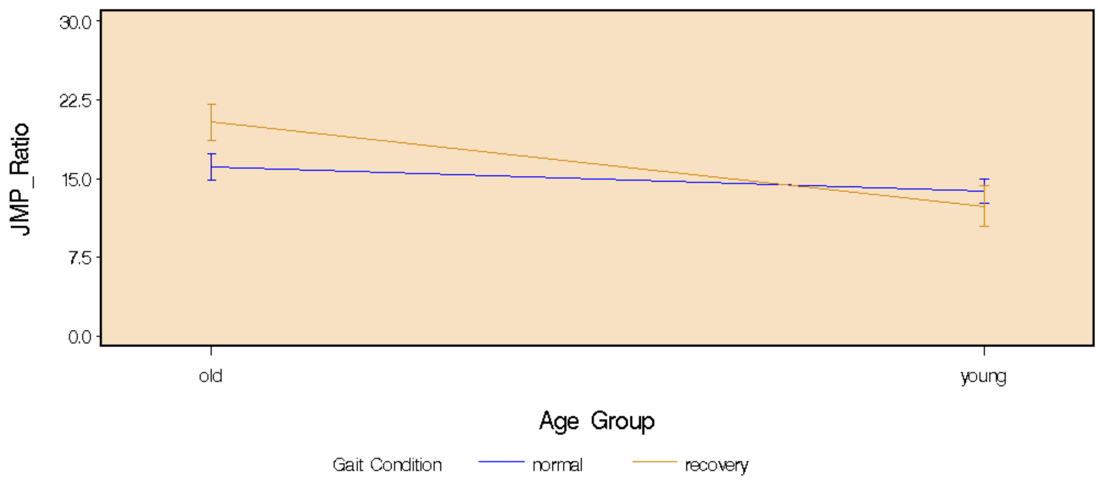

Figure 4.7-3 Interaction effect on ankle transverse JMP Ratio

4.7.4. Knee frontal plane

Table 4.7-4 Means and SD of knee frontal JMP Ratio (Percent)

	Old $\bar{X}(SD)$	Young $\bar{X}(SD)$
normal	36.01 (7.36)	39.11 (5.40)
reactive-recovery	39.49 (8.60)	22.73 (6.77)

There was a significant main effect of gait condition ($F(1,16) = 7.16$, $p = 0.017$) and a significant main effect of age group ($F(1,16.1) = 7.41$, $p = 0.015$). However, the interaction was also significant ($F(1,16) = 16.95$, $p < 0.001$).

Looking at the simple main effects, there was a significant effect of age group during reactive-recovery ($F(1,16) = 17.86$, $p < 0.001$), but not during normal walking ($F(1,15) = 1.04$, $p = 0.323$) showing that younger people possessed greater JMP Ratio than the elderly only during reactive-recovery condition. There was a significant effect of gait condition for the younger group ($F(1,16) = 29.11$, $p < 0.001$), but not for the elderly group ($F(1,15) = 0.85$, $p = 0.37$).

Therefore, younger people were found to have more knee frontal peak joint moment percentage than their old counterpart only during successful reactive-recovery. Meanwhile, only the younger group increased knee frontal peak joint moment percentage during reactive-recovery, with respect to normal walking. This is illustrated in Figure 4.7-4.

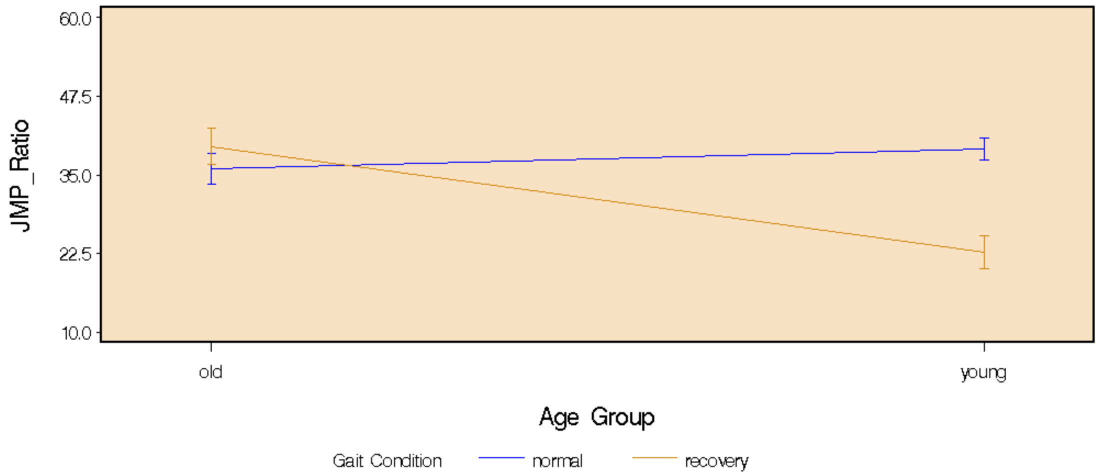

Figure 4.7-4 Interaction effect on knee frontal JMP Ratio

4.7.5. *Knee sagittal plane*

Table 4.7-5 Means and SD of knee transverse JMP Ratio (Percent)

	Old $\bar{X}\,(SD)$	Young $\bar{X}\,(SD)$
normal	45.22 (6.19)	46.89 (4.26)
reactive-recovery	36.49 (9.95)	71.13 (8.38)

There was a significant main effect of gait condition ($F(1,16) = 7.37$, $p = 0.015$) and a significant main effect of age group ($F(1,15.3) = 61.25$, $p < 0.001$). However, the interaction was also significant ($F(1,16) = 32.53$, $p < 0.001$).

Looking at the simple main effects, there was a significant effect of age group during reactive-recovery ($F(1,16) = 54.56$, $p < 0.001$), but not during normal walking ($F(1,15) = 0.41$, $p = 0.530$) showing that younger people possessed greater JMP Ratio than the elderly only during reactive-recovery condition. There was a significant effect of gait condition for both younger group ($F(1,16) = 52.04$, $p < 0.001$) and older group ($F(1,15) = 4.99$, $p = 0.040$).

Therefore, younger people were found to have more knee frontal peak joint moment percentage than their old counterpart only during successful reactive-recovery. Meanwhile, changes in JMP Ratio from reactive-recovery to normal walking were found to be different between young and old groups. Younger group increased knee sagittal peak joint moment percentage while older group decreased the percentage. This is illustrated in Figure 4.7-5.

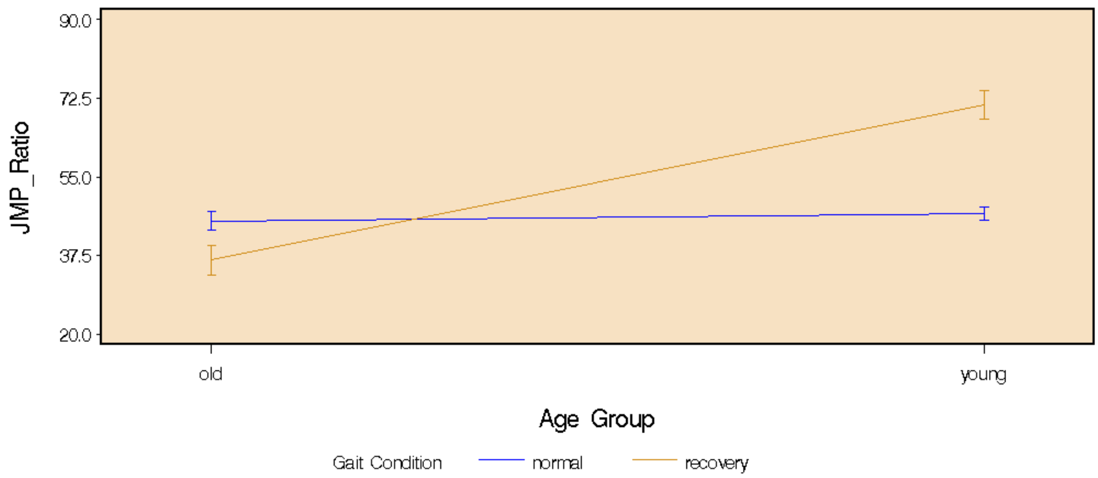

Figure 4.7-5 Interaction effect on knee sagittal JMP Ratio

4.7.6. *Knee transverse plane*

Table 4.7-6 Means and SD of knee transverse JMP Ratio (Percent)

	Old $\bar{X}\,(SD)$	Young $\bar{X}\,(SD)$
normal	18.78 (8.10)	13.33 (4.82)
reactive-recovery	24.02 (11.64)	10.12 (8.73)

There was a significant main effect of age group ($F(1,15.9) = 12.63$, $p = 0.003$) and no significant main effect of gait condition ($F(1,16) = 0.09$, $p = 0.770$). The interaction was not significant ($F(1,16) = 1.79$, $p = 0.199$).

Thus, the elderly group was found to have more knee transverse peak joint moment percentage than their younger counterparts, during both gait conditions. This is illustrated in Figure 4.7-6.

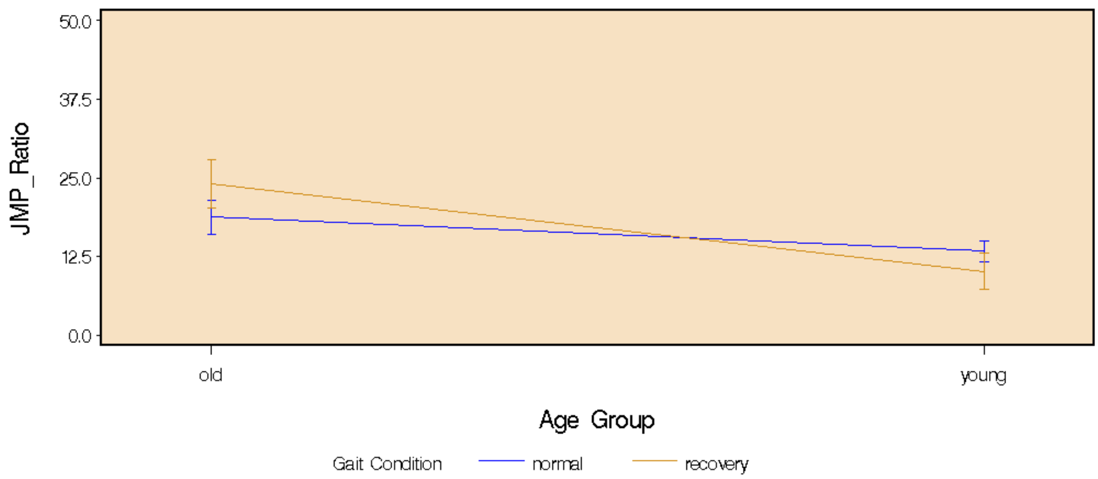

Figure 4.7-6 Interaction effect on knee transverse JMP Ratio

4.7.7. Hip frontal plane

Table 4.7-7 Means and SD of hip frontal JMP Ratio (Percent)

	Old $\bar{X}\,(SD)$	Young $\bar{X}\,(SD)$
normal	44.28 (2.43)	44.35 (2.89)
reactive-recovery	59.73 (7.26)	48.31 (5.89)

There was a significant main effect of gait condition (F(1,15.9) = 28.76, p < 0.001) and a significant main effect of age group (F(1,15.7) = 13.05, p = 0.002). However, the interaction was also significant (F(1,15.9) = 10.29, p = 0.006).

Looking at the simple main effects, there was a significant effect of age group during reactive-recovery (F(1,16) = 12.48, p = 0.003), but not during normal walking (F(1,15) < 0.01, p = 0.959) showing that younger people possessed greater JMP Ratio than the elderly only during reactive-recovery condition. There was a significant effect of gait condition for the elderly group (F(1,16) = 36.64, p < 0.001), but not for the younger group (F(1,15) = 3.22, p = 0.093).

Therefore, younger people were found to have less hip frontal peak joint moment percentage than their old counterpart only during successful reactive-recovery. Meanwhile, only the elderly increased their peak joint moment percentage in hip frontal plane during recovery condition, with respect to normal walking. This is illustrated in Figure 4.7-7.

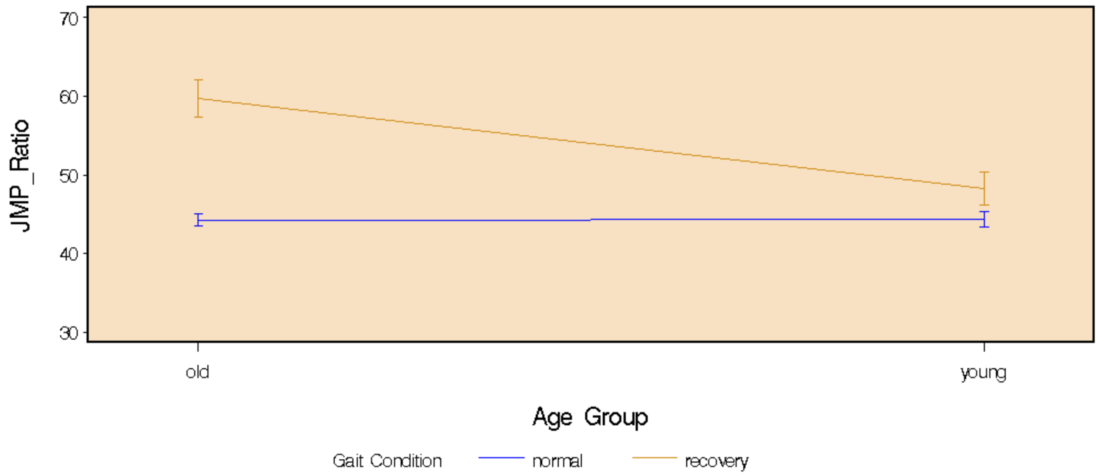

Figure 4.7-7 Interaction effect on hip frontal JMP Ratio

4.7.8. Hip sagittal plane

Table 4.7-8 Means and SD of hip sagittal JMP Ratio (Percent)

	Old $\bar{X}\,(SD)$	Young $\bar{X}\,(SD)$
normal	46.24 (2.59)	46.47 (4.57)
reactive-recovery	28.79 (5.48)	26.40 (8.92)

There was a significant main effect of gait condition (F(1,16) = 86.31, p < 0.001) and no significant main effect of age group (F(1,16) = 0.33, p = 0.573). The interaction was not significant (F(1,16) = 0.42, p = 0.525).

Hence, both age groups were found to reduce hip sagittal peak joint moment percentage during successful reactive-recovery, with respect to normal walking. This is illustrated in Figure 4.7-8.

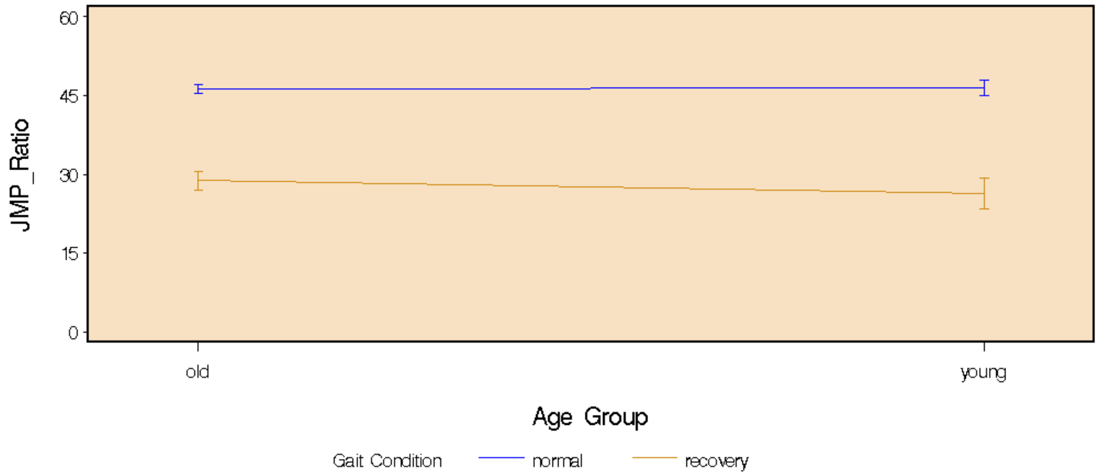

Figure 4.7-8 Interaction effect on hip sagittal JMP Ratio

4.7.9. Hip transverse plane

Table 4.7-9 Means and SD of hip transverse JMP Ratio (Percent)

	Old $\bar{X}\,(SD)$	Young $\bar{X}\,(SD)$
normal	9.48 (3.12)	7.52 (1.99)
reactive-recovery	11.48 (5.80)	23.01 (7.95)

There was a significant main effect of gait condition ($F(1,16) = 21.91$, $p < 0.001$) and a significant main effect of age group ($F(1,16) = 7.60$, $p = 0.014$). However, the interaction was also significant ($F(1,16) = 13.06$, $p = 0.002$).

Looking at the simple main effects, there was a significant effect of age group during reactive-recovery ($F(1,16) = 12.38$, $p = 0.003$), but not during normal walking ($F(1,15) = 2.31$, $p = 0.156$) showing that younger people possessed greater JMP Ratio than the elderly only during reactive-recovery condition. There was a significant effect of gait condition for the younger group ($F(1,16) = 26.60$, $p < 0.001$), but not for the elderly group ($F(1,15) = 0.83$, $p = 0.376$).

Therefore, younger people were found to have more hip transverse peak joint moment percentage than their old counterpart only during successful reactive-recovery.

Meanwhile, only the younger group increased hip transverse peak joint moment percentage during recovery condition, with respect to normal walking. This is illustrated in Figure 4.7-9.

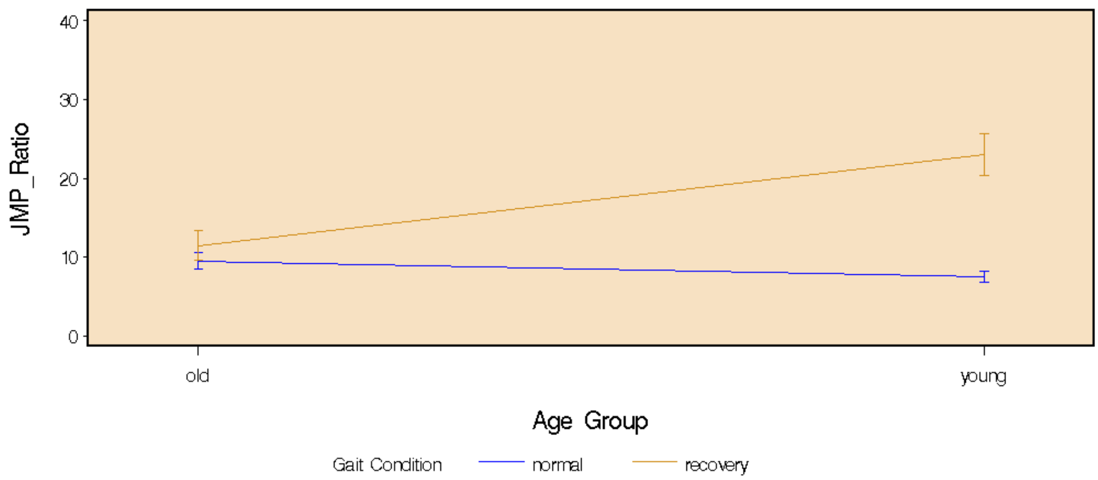

Figure 4.7-9 Interaction effect on hip transverse JMP Ratio

4.8. JMG Rate (joint moment generation rate)

JMG Rate in all the locations was analyzed using one-way between-subjects ANCOVA with two levels: young and old. Walking velocity was considered as a covariate. Descriptive summary of JMG Rate was shown in Table 4.8-1. Hip sagittal plane location was excluded from the analysis because JMG Rate concept was incompatible with the unique double-peak joint moment profile detected in hip sagittal plane.

At the ankle joint, there was a significant effect of age group in the sagittal plane ($F(1,16) = 6.68$, $p = 0.020$), and no significant effects of age group in the frontal plane ($F(1,16) = 0.23$, $p = 0.636$) or in the transverse plane ($F(1,16) = 0.36$, $p = 0.559$), showing that the younger group produced higher JMG Rate than the elderly only in ankle sagittal plane.

At the knee joint, there was a significant effect of age group in the sagittal plane ($F(1,16) = 9.64$, $p = 0.007$), and no significant effects of age group in the frontal plane ($F(1,16) = 0.21$, $p = 0.656$) or in the transverse plane ($F(1,16) = 1.71$, $p = 0.210$), showing that the younger group produced higher JMG Rate than the elderly only in knee sagittal plane.

Table 4.8-1 Means and SD of JMG Rate (N*m / kg*sec)

		Old $\bar{X}\,(SD)$	Young $\bar{X}\,(SD)$	Effect
Ankle	Frontal	3.59 (1.58)	3.17 (2.05)	
	Sagittal	4.01 (2.62)	8.43 (4.41)	*
	Transverse	2.01 (1.09)	1.71 (1.10)	
Knee	Frontal	3.47 (1.81)	3.15 (0.83)	
	Sagittal	3.08 (1.41)	7.54 (4.07)	*
	Transverse	1.82 (0.90)	1.23 (1.03)	
Hip	Frontal	6.23 (1.59)	5.73 (1.43)	
	Transverse	1.03 (0.57)	1.98 (0.82)	*

* indicate significant aging effect

At the hip joint, there was a significant effect of age group in the transverse plane ($F(1,16) = 7.91$, $p = 0.013$), and no significant effects of age group in the frontal plane ($F(1,16) = 0.48$, $p = 0.496$), showing that the younger group produced higher JMG Rate than the elderly only in hip transverse plane.

Therefore, in the successful recovery process, the younger group was found to generate joint moment faster than their older counterparts during the period from joint moment activation to peak. However, such significant aging effect was only evident in ankle sagittal plane, knee sagittal plane, and hip frontal plane.

4.9. JMP Power (peak joint power magnitude)

JMP Power in all locations (joint*reference plane) was analyzed using two-way mixed-subjects ANCOVA with four conditions: old/normal, old/recovery, young/normal, young/recovery. Walking velocity was considered as a covariate.

4.9.1. *Ankle frontal plane*

Table 4.9-1 Means and SD of ankle frontal JMP Power (J/kg)

	Old $\bar{X}\,(SD)$	Young $\bar{X}\,(SD)$
normal	0.22 (0.15)	0.35 (0.35)
reactive-recovery	1.31 (0.93)	1.45 (0.87)

There was a significant main effect of gait condition ($F(1,10.1) = 21.59$, $p < 0.001$) and no significant main effect of age group ($F(1,9.9) = 0.47$, $p = 0.509$). The interaction was not significant ($F(1,10.1) < 0.01$, $p = 0.984$).

Thus, both young and old groups generated greater peak joint powers in ankle frontal plane during successful reactive-recovery than during normal walking. This is illustrated in Figure 4.9-1.

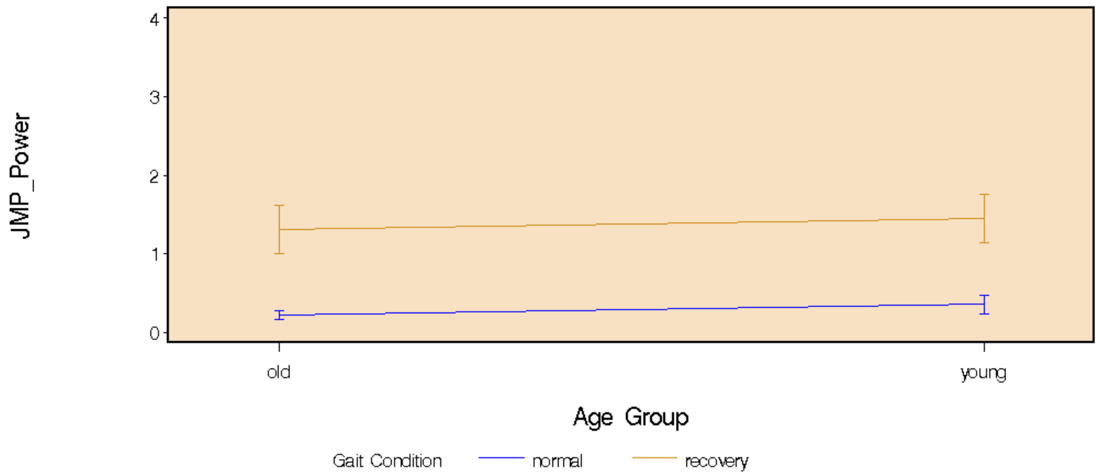

Figure 4.9-1 Interaction effect on ankle frontal JMP Power

4.9.2. *Ankle sagittal plane*

Table 4.9-2 Means and SD of ankle sagittal JMP Power (J/kg)

	Old $\bar{X}\,(SD)$	Young $\bar{X}\,(SD)$
normal	4.36 (0.84)	5.15 (0.79)
reactive-recovery	7.62 (9.68)	8.53 (6.22)

There was no significant main effect of gait condition (F(1,15.7) = 2.73, p = 0.119) and no significant main effect of age group (F(1,15.7) = 0.18, p = 0.675). The interaction was not significant (F(1,15.7) < 0.01, p = 0.973).

Hence, the peak joint power produced by both age groups was found to be not different during both normal walking and recovery conditions. This is illustrated in Figure 4.9-2.

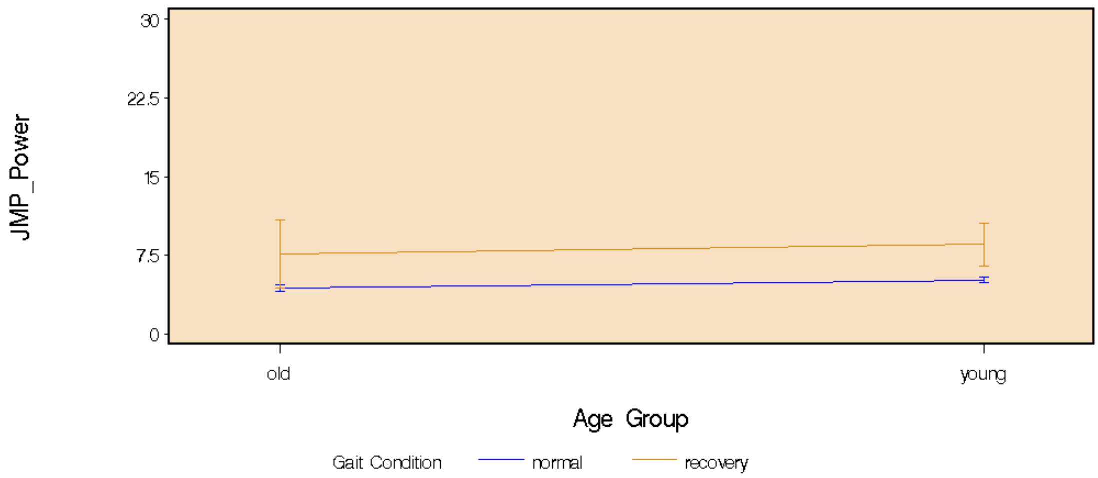

Figure 4.9-2 Interaction effect on ankle sagittal JMP Power

4.9.3. Ankle transverse plane

Table 4.9-3 Means and SD of ankle transverse JMP Power (J/kg)

	Old $\bar{X}\,(SD)$	Young $\bar{X}\,(SD)$
normal	0.21 (0.10)	0.23 (0.15)
reactive-recovery	1.37 (1.01)	1.38 (0.71)

There was a significant main effect of gait condition (F(1,15.4) = 31.80, p < 0.001) and no significant main effect of age group (F(1,15.5) < 0.01, p = 0.943). The interaction was not significant (F(1,15.4) < 0.01, p = 0.977).

Thus, both young and old groups generated greater peak joint power in ankle transverse plane during successful reactive-recovery, with respect to normal walking. This is illustrated in Figure 4.9-3.

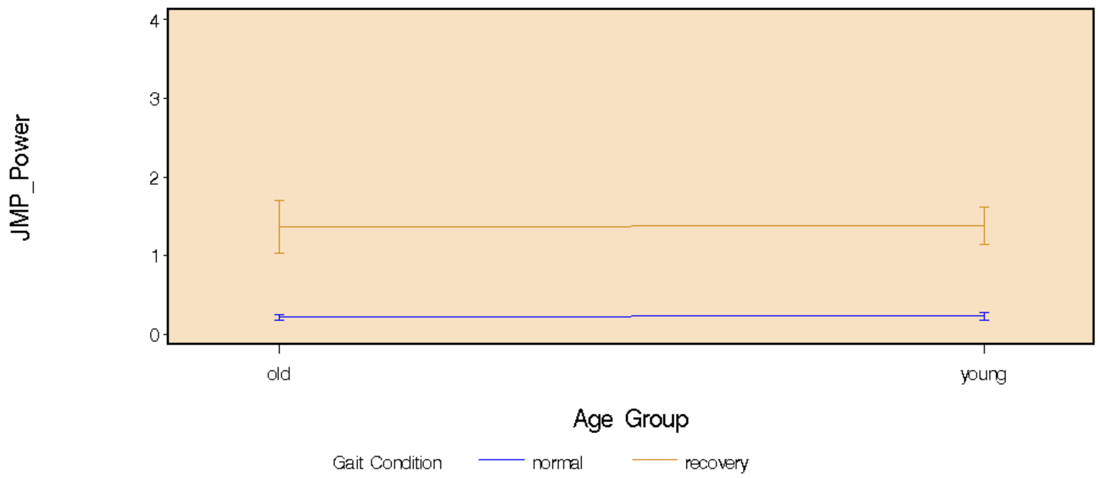

Figure 4.9-3 Interaction effect on ankle transverse JMP Power

4.9.4. Knee frontal plane

Table 4.9-4 Means and SD of knee frontal JMP Power (J/kg)

	Old $\bar{X}\,(SD)$	Young $\bar{X}\,(SD)$
normal	0.17 (0.07)	0.23 (0.12)
reactive-recovery	0.71 (0.35)	1.25 (0.48)

There was a significant main effect of gait condition ($F(1,16.2) = 66.03$, $p < 0.001$) and a significant main effect of age group ($F(1,17.1) = 7.17$, $p = 0.016$). However, the interaction was also significant ($F(1,16.2) = 6.45$, $p = 0.022$).

Looking at the simple main effects, there was a significant effect of gait condition for the younger group ($F(1,14) = 20.91$, $p < 0.001$) as well as the older group ($F(1,14) = 33.75$, $p < 0.001$). There was a significant effect of age group during reactive-recovery ($F(1,13) = 6.17$, $p = 0.027$), but not during normal walking ($F(1,15) = 1.83$, $p = 0.196$).

Thus, both young and old groups generated greater peak joint power in knee frontal plane during successful reactive-recovery, with respect to normal walking. Meanwhile, the younger group was found to develop greater peak joint power than their older counterparts only during reactive-recovery. This is illustrated in Figure 4.9-4.

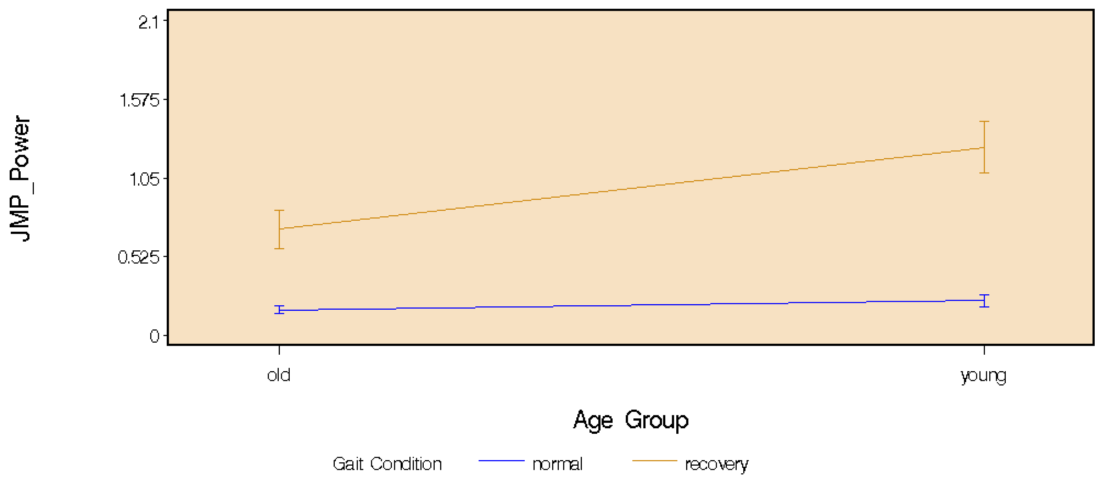

Figure 4.9-4 Interaction effect on knee frontal JMP Power

4.9.5. Knee sagittal plane

Table 4.9-5 Means and SD of knee sagittal JMP Power (J/kg)

	Old $\bar{X}\,(SD)$	Young $\bar{X}\,(SD)$
normal	2.46 (1.54)	2.39 (1.13)
reactive-recovery	3.62 (3.22)	8.47 (6.55)

There was a significant main effect of gait condition ($F(1,16) = 8.39$, $p = 0.011$) and no significant main effect of age group ($F(1,16) = 3.57$, $p = 0.077$). The interaction was not significant ($F(1,16) = 3.86$, $p = 0.067$).

Thus, both young and old groups generated greater peak joint power in knee sagittal plane during successful reactive-recovery, with respect to normal walking. This is illustrated in Figure 4.9-5.

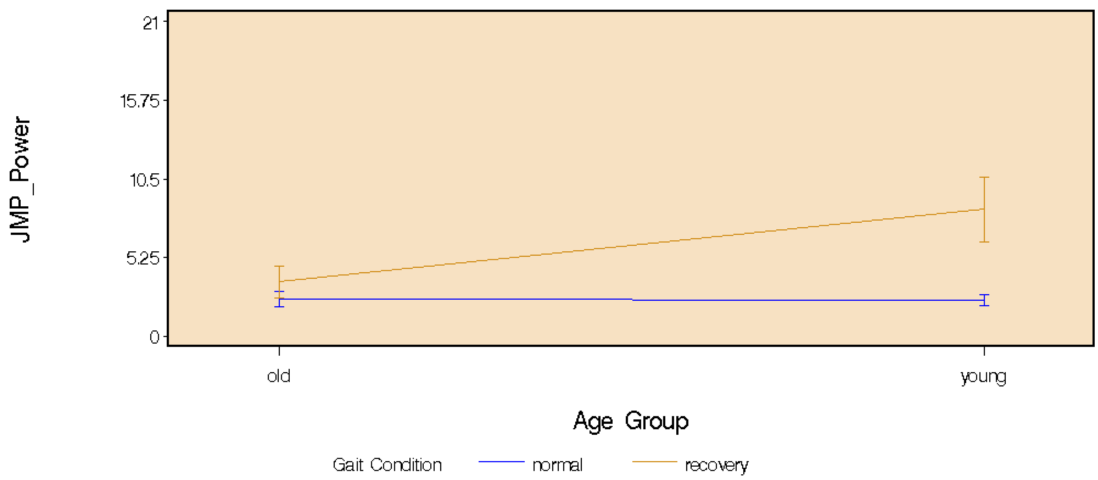

Figure 4.9-5 Interaction effect on knee sagittal JMP Power

4.9.6. Knee transverse plane

Table 4.9-6 Means and SD of knee transverse JMP Power (J/kg)

	Old $\bar{X}\,(SD)$	Young $\bar{X}\,(SD)$
normal	0.33 (0.32)	0.28 (0.24)
reactive-recovery	1.72 (0.85)	1.10 (0.55)

There was a significant main effect of gait condition (F(1,16) = 29.52, p < 0.001) and no significant main effect of age group (F(1,16) = 4.50, p = 0.050). The interaction was not significant (F(1,16) = 1.96, p = 0.181).

Thus, both young and old groups generated greater peak joint power in knee transverse plane during successful reactive-recovery, with respect to normal walking. This is illustrated in Figure 4.9-6.

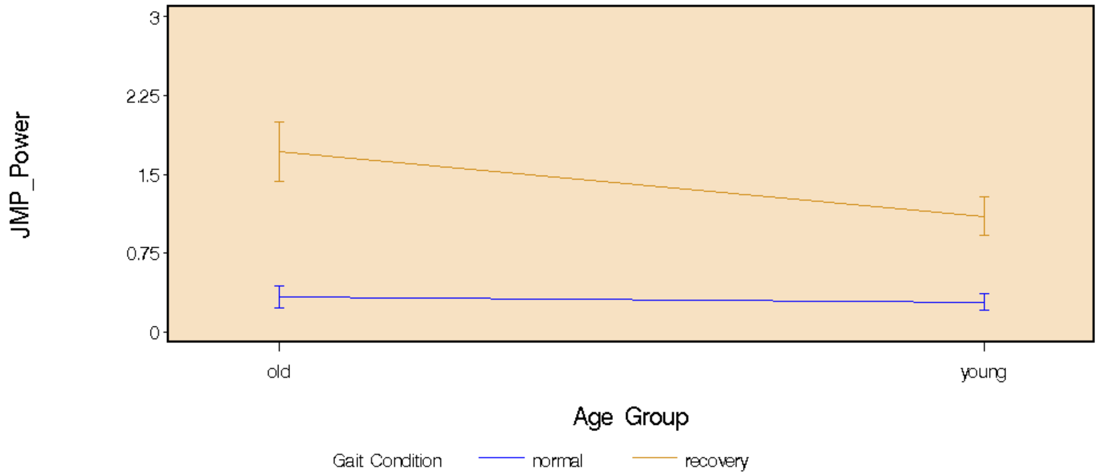

Figure 4.9-6 Interaction effect on knee transverse JMP Power

4.9.7. Hip frontal plane

Table 4.9-7 Means and SD of hip frontal JMP Power (J/kg)

	Old $\bar{X}\,(SD)$	Young $\bar{X}\,(SD)$
normal	0.56 (0.22)	0.56 (0.27)
reactive-recovery	1.28 (0.78)	1.17 (0.52)

There was a significant main effect of gait condition ($F(1,9.15) = 16.84$, $p = 0.003$) and no significant main effect of age group ($F(1,9.27) = 0.14$, $p = 0.722$). The interaction was not significant ($F(1,9.15) = 0.07$, $p = 0.792$).

Thus, both young and old groups generated greater peak joint power in hip frontal plane during successful reactive-recovery, with respect to normal walking. This is illustrated in Figure 4.9-7.

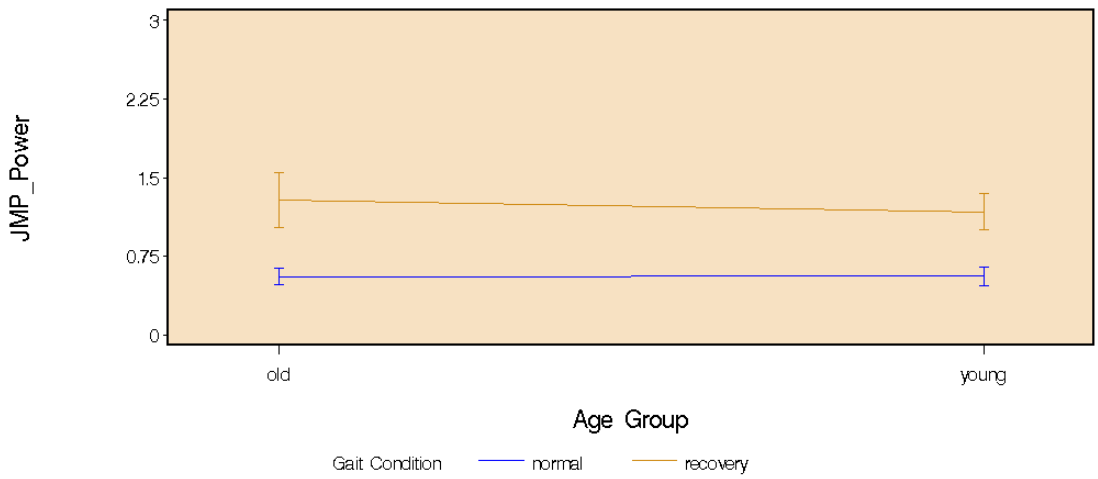

Figure 4.9-7 Interaction effect on hip frontal JMP Power

4.9.8. Hip sagittal plane

Table 4.9-8 Means and SD of hip sagittal JMP Power (J/kg)

	Old $\bar{X}\,(SD)$	Young $\bar{X}\,(SD)$
normal	2.48 (0.18)	2.82 (0.99)
reactive-recovery	1.94 (0.55)	2.60 (1.09)

There was no significant main effect of gait condition ($F(1,8.87) = 1.21$, $p = 0.301$) and no significant main effect of age group ($F(1,10.7) = 3.06$, $p = 0.109$). The interaction was not significant ($F(1,8.87) = 0.64$, $p = 0.445$).

Hence, the peak joint power produced by both age groups was found to be not different during both normal walking and recovery conditions. This is illustrated in Figure 4.9-8.

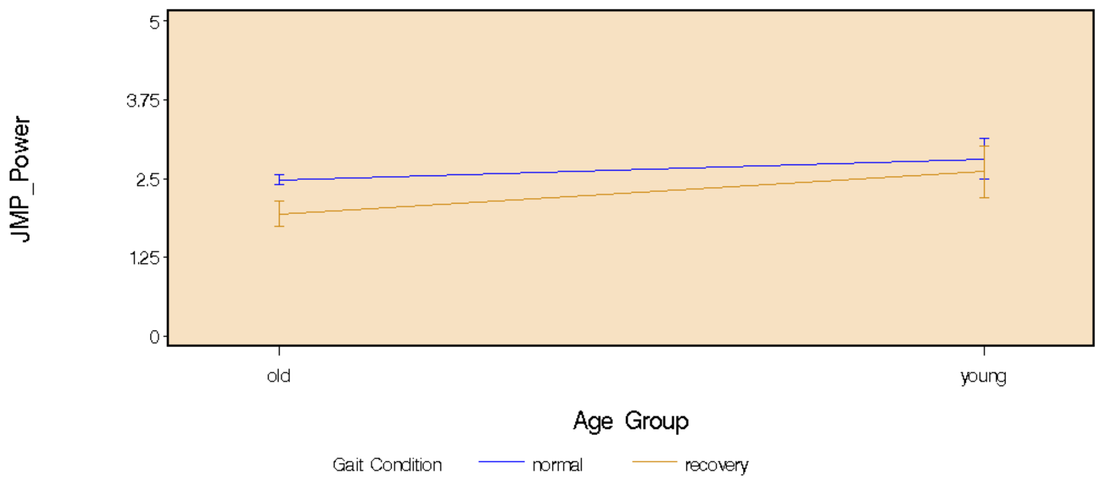

Figure 4.9-8 Interaction effect on hip sagittal JMP Power

4.9.9. Hip transverse plane

Table 4.9-9 Means and SD of hip transverse JMP Power (J/kg)

	Old $\bar{X}\,(SD)$	Young $\bar{X}\,(SD)$
normal	0.17 (0.06)	0.13 (0.05)
reactive-recovery	0.33 (0.21)	0.74 (0.47)

There was a significant main effect of gait condition (F(1,13.9) = 17.82, p < 0.001) and no significant main effect of age group (F(1,14.2) = 3.55, p = 0.080). However, the interaction was significant (F(1,13.9) = 6.55, p = 0.023).

Looking at the simple main effects, there was a significant effect of gait condition for younger group (F(1,14) = 13.29, p = 0.002), but not elderly group (F(1,16) = 3.74, p = 0.074), showing that younger people generated greater peak joint powers in both gait conditions.

Therefore, only younger people generated greater peak joint power in hip transverse plane during successful reactive-recovery, with respect to normal walking. This is illustrated in Figure 4.9-9.

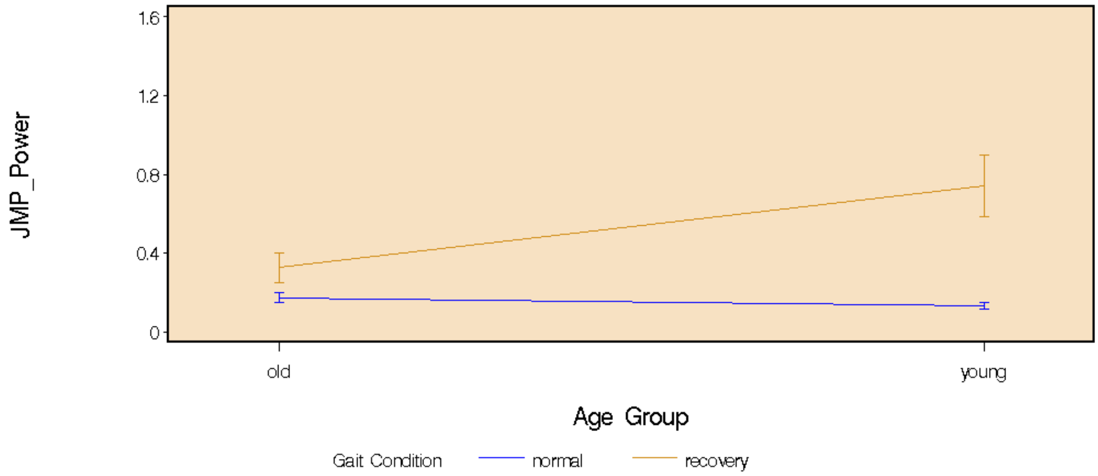

Figure 4.9-9 Interaction effect on hip transverse JMP Power

4.10. Strength measurement correlation analysis

Bivariate correlation analyses were used to determine the relationship among three dynamic levels ($30^\circ \cdot s^{-1}$, $60^\circ \cdot s^{-1}$ and $120^\circ \cdot s^{-1}$) of maximum isokinetic joint torque measurements at ankle and knee joints. Measurement at dynamic level of $120^\circ \cdot s^{-1}$ was adopted in current study was based on the instant ankle angular velocity at heel contact. Summary of correlation analysis results was listed in Table 4.10-1.

The results indicated statistically significant relationship ($p < 0.001$) between $30^\circ \cdot s^{-1}$ and $60^\circ \cdot s^{-1}$ at ankle joint ($r = 0.91$) and knee joint ($r = 0.98$). In other words, the significant positive relationship indicated that isokinetic strength at $30^\circ \cdot s^{-1}$ was higher related to the higher isokinetic strength at $60^\circ \cdot s^{-1}$ at ankle joint and knee joint.

The results indicated statistically significant relationship ($p < 0.001$) between $60^\circ \cdot s^{-1}$ and $120^\circ \cdot s^{-1}$ at ankle joint ($r = 0.88$) and knee joint ($r = 0.96$). In other words, the significant positive relationship indicated that isokinetic strength at $60^\circ \cdot s^{-1}$ was higher related to the higher isokinetic strength at $120^\circ \cdot s^{-1}$ at ankle joint and knee joint.

Table 4.10-1 Summary of strength correlation analysis

Joint	Dynamic level relationships	r	p	Significant effect
Ankle	$30^\circ \cdot s^{-1}$ and $60^\circ \cdot s^{-1}$	0.91	<0.001	*
	$30^\circ \cdot s^{-1}$ and $120^\circ \cdot s^{-1}$	0.84	<0.001	*
	$60^\circ \cdot s^{-1}$ and $120^\circ \cdot s^{-1}$	0.88	<0.001	*
Knee	$30^\circ \cdot s^{-1}$ and $60^\circ \cdot s^{-1}$	0.98	<0.001	*
	$30^\circ \cdot s^{-1}$ and $120^\circ \cdot s^{-1}$	0.94	<0.001	*
	$60^\circ \cdot s^{-1}$ and $120^\circ \cdot s^{-1}$	0.96	<0.001	*

r indicate Pearson correlation coefficient

* indicate significant relationship between variables

The results indicated statistically significant relationship ($p < 0.001$) between $30^\circ \cdot s^{-1}$ and $120^\circ \cdot s^{-1}$ at ankle joint ($r = 0.84$) and knee joint ($r = 0.94$). In other words, the significant positive relationship indicated that isokinetic strength at $30^\circ \cdot s^{-1}$ was higher related to the higher isokinetic strength at $120^\circ \cdot s^{-1}$ at ankle joint and knee joint.

4.11. Walking velocity and step length

Walking velocity and step length in normal gait were analyzed using one-way between-subjects ANOVA with two levels: 2002 and 2003. Descriptive summary of JMA Time was shown in Table 4.11-1. There are 20 young and 20 old participants recruited for the experiments conducted in 2002, and 10 young and 10 old participants in 2003.

There was no significant effect of experiment year on walking velocity ($F(1,53) = 3.12$, $p = 0.083$), neither on step length ($F(1,53) = 0.1053$, $p = 0.747$), confirming the identical nature of experiments conducted in 2002 and 2003.

Table 4.11-1 Mean and SD of walking velocity and step length

	2002 $\bar{X}\,(SD)$	2003 $\bar{X}\,(SD)$
Walking Velocity (mm/s)	1299.29 (182.57)	1206.33 (184.39)
Step Length (mm)	682.53 (62.70)	688.85 (77.38)

Chapter 5

5. DISCUSSION AND CONCLUSIONS

5.1. Hypotheses and experimental findings

5.1.1. Hypothesis (a) - "The elderly will have longer JMA Time (ankle, knee and hip) than the younger individuals during successful reactive-recovery."

JMA Time (joint moment activation time) was determined as the interval from heel contact until the reactive joint moment threshold. This variable was expected to reveal the characteristics of reactive joint moment initiation during successful reactive-recovery.

The time available to make appropriate postural adjustment was regarded as one important factor to recover and maintain dynamic balance in the event of unexpected external balance disturbance such as slip/fall accidents. Numerous studies have documented the age-related decline of postural control (Sheldon, 1963; Woolacott, Sumway-Cook, & Nashner, 1982). This decline in postural control was also believed to be closely related with a higher risk of falling (Pyykko, Jantti, & Aalto, 1990; Alexander, Shephard, Mian, & Schultz, 1992; Brocklehurst, Robertson, & James-Groom, 1982; Overstall, Exton-Smith, Imms, & Johnson, 1977). Robinovitch et al. (2002) investigated the reactive joint moment development on balance recovery via lean-release experiments. They concluded that human ability to recover balance following an unexpected perturbation is limited substantially by the available reaction time.

No aging effect on JMA Time was found during reactive-recovery in current study. This result suggested joint moment activation time as a common characteristic for all the individuals who were able to achieve successful balance recovery. The joint moment activation time (in average 271ms) found in current study was incomparable to the timing of responses reported from other postural balance studies. During the initial 150ms following the applied external perturbation, Runge et al. (1998) observed only passive joint movements. However, active lower extremity joint moments became evident at about 150-200ms in their study. Robinovitch et al (2002) found even shorter reaction times (in average 100ms). Such discrepancies in reaction time may be the result of two aspects. First, the experimental protocol in current study was different from the one conducted by Robinovitch et al in 2002. Though the nature of both experiments focused on the balance recovery from unexpected external perturbation, the perturbation created in current study was walking on slippery surface while Robinovitch et al. (2002) adopted the forward lean-release approach. Cham et al (2001) also concluded that reactive joint moment responses resulting from slip/fall experiment were different from the results in simulated slip event (unexpected forward transitional perturbation) studied by Ferber (2002) and reported an average joint moment reaction between 190 and 350ms which was in close agreement with current findings. Such increases in reaction time during slip/fall can be due to the typical delay of slip initiation, which usually occurred at about 70-150 ms after heel contact (Perkins, 1978). Second, the definition of reactive joint moment onset was different among studies. The current study defined the joint moment reaction threshold (Mean +/- 2SD) based on the joint moment history produced during slip period, while Robinovitch et al. (2002) defined the reactive joint moment onset as the average joint moment plus 5 Nm.

In summary, the joint moment activation time was suggested to be a common characteristic for all the individuals who were able to achieve successful reactive-recovery, due to the fact that aging effect was not present on JMA Time.

5.1.2. Hypothesis (b) - "The elderly will have longer JMAP Time (as of ankle, knee and hip) than the younger individuals during successful reactive-recovery."

JMAP Time (joint moment activation to peak time) was determined as the interval from joint moment activation threshold to peak joint moment. This variable was expected to reveal the characteristics of reactive joint moment development during successful reactive-recovery.

Being able to rapidly develop peak joint moment was suggested to be critical to balance recovery by previous research. Thelen et al (1996) studied aging effect on ankle dorsiflexion and plantarflexion joint moment development during rapid isokinetic exertions. They found elderly individuals required more time to reach maximum joint torque, which indicated the declines in the ability of healthy old adults to rapidly develop ankle sagittal joint moments. A prolonged time to peak torque could indicate reduced recruitment of type II fibers (Kannus, 1994). Furthermore, it was supported by epidemiological evidence that risk for falls increases with the increases in the time required to reach peak joint moment (Lord et al., 1994; Nevitt et al., 1991).

No aging effect on JMAP Time was found for all the locations. This result suggested JMAP time to be another common characteristic possessed by all the individuals who were able to achieve successful reactive-recovery. Consider the above JMA Time related findings, it could be inferred that in order to reduce the risks for fall, the elderly has to be as rapid as their younger counterparts in terms of initiating and developing reactive joint moment.

In summary, JMAP Time was suggested as a common characteristic for all the individuals who were able to achieve successful reactive-recovery.

5.1.3. Hypothesis (c) - "JMP Magnitude (as of ankle, knee and hip) will be larger during successful reactive-recovery than during normal walking for both young and old."

Characteristic stance phase joint moment profiles were observed in both normal walking and reactive-recovery. The average joint moment profiles in normal gait from the current study were in general agreement with previous literature (Besier, Sturnieks, Alderson, & Lloyd, 2003; Allard et al., 1996; Eng et al., 1995; Ounpuu et al., 1996; Glitsch et al., 1997; Alkjaer et al., 2001; Cham et al., 2001; Silva & Ambrosio, 2003; Redfern & DiPasquale, 1997). Contrary to normal gait, joint moment magnitudes in reactive-recovery were found to be highly variable. Such large variability was mainly contributed to the individual-specific nature of slip/fall accidents, which was also evident in a previous study (Cham et al., 2001).

Reactive-recovery results from current study were considerably different from previous literature. Cham et al. (2001) performed 2D sagittal joint moment analysis on three conditions: normal gait, successful reactive-recovery and unsuccessful reactive-recovery (fall). Their findings about the joint moment profiles during successful reactive-recovery are summarized below:

- Ankle joint was characterized by a plantarflexor dominant joint moment which is similar to those computed in normal gait;
- A considerable bias (relative to normal gait) towards flexion moment was observed at the knee between 25% and 45% into stance;
- The hip moment was governed by a large extensor during that same time.

Nevertheless, reactive-recovery joint moment profiles obtained from the current study were characterized by

- A significantly large dorsiflexor ankle sagittal joint moment;
- A extensor dominant knee sagittal joint moment; and
- A double-peak extensor hip sagittal joint moment.

Furthermore, on average, sagittal joint moments for the younger group in the current study reached as high as 3.66, 3.33, and 1.28 (Nm/kg) for ankle, knee and hip joints, respectively. These joint moment peaks are far beyond the previous findings, which was less than 1 in average.

Such discrepancy in reactive-recovery joint moment profiles may be mainly contributed to the different time interval adopted in the different studies. In the current study, the targeting time interval (about 600 ms) was from heel contact to toe off, which covered the whole balance recovery process. In previous studies, the targeting time interval was limited to 55% from heel contact to toe off. As evident in ankle, knee and hip joint moment profiles (Figure 1,2,3) from the current study, major reactive joint moments were developed during second half of targeting time interval (50% to 100% of stance). As suggested by Rogers et al. (2003) and Do et al. (1982) in their investigation of protective stepping, two phases in balance recovery were identified: 1) initial voluntary reaction phase and 2) controlled reactive action. It can be then plausibly assumed that first time interval (up to 55% from heel contact to toe off) falls into the voluntary reaction, while the second time interval which was elaborated in current study might reflect the controlled reactive-recovery strategy. Thus, it can be suggested that the targeting time interval has to be extended to the whole stance phase if the controlled recovery process rather than voluntary initiation process was desired.

A significant gait condition effect on JMP Magnitude was found in almost all the locations (except knee transverse and hip transverse planes). Both age groups developed significant higher peak joint moments in reactive-recovery than in normal walking. This finding confirmed the proposed hypothesis. Such increases in peak joint moments indicated that considerable higher muscle activations were required for an individual to successful recover their balance from unexpected slips, which was also supported by other postural control studies (Chandler, Duncan, & Studenski, 1990; Wojcik, Thelen, Schultz, Ashton-Miller, & Alexander, 1999). In another word, inability to generate required joint moments might contribute to the slip-induced fall accidents. Instead of increasing peak joint moments to compensate for balance recovery, hip sagittal peak joint moments were found to be significantly reduced. It was documented that the two major functions of hip joint was forward movement propulsion, which was accomplished mainly by sagittal plane joint moment, and balance maintenance of the upper body, which was accomplished mainly by frontal plane joint moment (Winter, 1983). Evident by the increase of hip frontal joint moment and decrease of hip sagittal joint moment, it could be inferred that during reactive-recovery, the hip joint was mainly involved with the balance maintenance of upper body. The above findings may be summarized into the joint moment generation strategy responding to the slip-induced perturbations: ankle and knee joints actively served as the major actuator for balance recovery by producing considerable larger joint moments while hip joint passively maintain upright posture for upper body by decreasing peak joint moment generation.

A significant aging effect on JMP Magnitude was found only in ankle sagittal, knee sagittal and knee transverse planes. At these locations, the peak joint moments developed by the younger group were significantly higher than those developed by the older group. Such JMP Magnitude differences were mainly due to the distinct joint moment distribution strategies adopted by different age groups. According to the findings about JMP Ratio (Table 4.7-1~9), the elderly relied more on the frontal joint moment instead of sagittal joint moment at the ankle joint. At the knee joint, the elderly relied more on the frontal and transverse joint moment for balance recovery.

In summary, the proposed hypothesis was confirmed that significant higher peak joint moments were detected in most of the locations during reactive-recovery. The ankle and knee joints were assumed major active joints in balance recovery while the hip joint was suggested to passively maintain the balance of upper body. Due to the distinct joint moment distribution strategies adopted, higher sagittal peak joint moments generated by the younger group than the older group were detected at the ankle and knee joints.

5.1.4. Hypothesis (d) - "The elderly will have larger JMG Ratio (as of ankle and knee) than the younger individuals during successful reactive-recovery."

JMG Ratio (joint moment generation ratio) determined as ratio of the peak joint moment to the strength measurement at the same location, was expected to reveal the utilization of lower extremity muscle strength capacity in generating joint moment.

Peak joint moments were found to be far beyond their corresponding strength capacity during reactive-recovery. This was particularly true for the ankle joint moments for both young (on average: 836% of isokinetic measurement) and old (on average: 634% isokinetic measurement) participants. Similar findings were also observed in previous studies in balance recovery after tripping (Pijnappels, Bobbert, & van Dieen, 2004), in which peak ankle joint moment was found to exceed the isometric capacity of elderly females by a factor of about 2.4. In current study, it was a limitation to compare the data derived using dynamometers directly with the estimates of joint moments from kinematics due to numerous factors including angular velocity, subject characteristics as well as the fact that voluntary activation in isokinetic conditions is not necessarily the maximal (Pijnappels et al., 2004). Nevertheless, such extreme JMG Ratios still indicated the requirements of high joint moments in reactive-recovery process.

Increased JMG Ratios during reactive-recovery over normal walking for both age groups were evident. Such significant increases in JMG Ratio were mainly due to the increase in peak joint moment during reactive-recovery, given the same strength measurements used in both gait conditions. As suggested by previous studies (Larsson, Grimby, & Karlsson, 1979; Wolfson, Whipple, Amerman, Kaplan, & Kleinberg, 1985; Woolacott, 1986; Bonder & Wagner, 1994; Campbell, Borrie, & Spears, 1989), declines in muscle strength capacity and muscle force production have an important effect on initiation and recovery of slip and fall accidents. Because the strength measurement served as an indicator of individual's muscle strength capability, it could be inferred that successful reactive-recovery required significant higher muscle activity to produce sufficient joint moments, in terms of individual's muscle strength capacity. In other words, inability to satisfy such higher joint moment generation demand due to declined muscle strength capacity might be one important factor leading to unsuccessful reactive-recovery, namely slip-induced falls. Future investigation incorporating falls experimental data will be necessary to confirm this proposition.

No aging effect on JMG Ratio was found in both reactive-recovery and normal gait conditions. Similar to the results related to JMA and JMAP timing variables, consistent increases in JMG Ratio during reactive-recovery might suggest JMG Ratio as a common characteristic for all the individuals (including the young and elderly groups) who were able to achieve successful reactive-recovery. Such common characteristic proposed the feasibility to increase the likelihood of successful reactive-recovery by elevating the strength capability of the muscle groups surrounding ankle and knee joints.

In summary, the increases in JMG Ratio during reactive-recovery indicated the significant higher muscle force production requirement relative to the available muscle strength capability for both young (in average: 836% at ankle and 321% at knee) and old (in average: 634% at ankle and 177% at knee) participants. Meanwhile, the consistent increase in JMG Ratio implied JMG Ratio as a common characteristic for all the individuals who were able to recover successfully from slip-induced fall. Lower extremity strength training, especially targeting ankle and knee joints, is expected to increase the likelihood of balance recovery from unexpected slips.

5.1.5. *Hypothesis (e) - "The sagittal plane JMP Ratio (as of ankle, knee and hip) for each lower extremity joint will be smaller during reactive-recovery than during normal walking for both age groups."*

JMP Ratio (peak joint moment ratio) determined as the percentage of sagittal peak joint moment in term of total joint moment in all three reference planes, was used to describe the joint moment distributions in three reference planes.

The proposed hypothesis was confirmed for the elderly group. Significant decreases in sagittal JMP Ratio generated at ankle, knee and hip joints were observed during reactive-recovery. To compensate for the decrease in sagittal plane, JMP Ratio was significantly increased in the frontal plane at ankle (41.16%) and hip (59.73%) joints. Such JMP Ratio increase promoted ankle frontal and hip frontal plane joint moments to the dominant role in balance recovery. Previous studies have ankle and hip frontal joint moments as the two levels of control to fine-tune the medial acceleration of the whole body center-of-mass (Mackinnon et al., 1993). It has been identified that total body medial/lateral balance is achieved primarily by the supporting foot. Meanwhile, Control of HAT (head, arms and trunk segment) about the supporting hip is achieved by the hip abductors. Considering the findings from the current study, it can be concluded that the reactive joint moment strategies adopted by the elderly were to primarily correct body frontal plane balance. Adoption of such strategies might be plausibly explained by the limitation in available muscle strength forcing the elderly to activate adjacent muscle groups to assist in balance recovery, given the fact that peak joint moment observed in ankle and knee joints were no more than 2 Nm/kg in reactive-recovery. Current findings also justified the application of 3D joint moment analysis in the condition that complete reactive-recovery information in slip/fall accidents was preferred.

Distinct joint moment distribution strategies were adopted by the younger group during reactive-recovery. JMP Ratios for the younger group showed a significant decrease in hip sagittal moments, a significant increase in knee sagittal moments and no change in ankle sagittal moments during reactive-recovery compared to normal walking. To compensate for the changes in knee and hip sagittal planes, JMP Ratios were decreased in the knee frontal plane and increased in the hip transverse plane. Thus, it can be inferred that the younger group still highly relied on the ankle and knee sagittal joint moment to achieve balance recovery. Adoption of such strategy might be plausibly explained by the insufficiencies of available muscle strength to enable further increasing sagittal joint moment production, given the fact that peak joint moment observed in ankle and knee sagittal plane was in average 3.66 and 3.33, respectively.

An aging effect on JMP Ratio was found to support the above mentioned joint moment distribution strategy. Sagittal JMP Ratio at ankle and knee joints was significantly higher for the younger group than their older counterparts during reactive-recovery. Accordingly, the elderly produced significantly higher JMP Ratios than their younger counterparts, namely ankle transverse, knee frontal, knee transverse, and hip frontal planes, during reactive-recovery. Another finding to be noticed was that in normal walking, the elderly were found to have higher JMP Ratios in the ankle transverse and knee transverse planes. This finding indicated the increased transverse balance maintenance needs associated with aging.

In summary, distinct joint moment distribution strategies were adopted by different age groups, which may be due to the possible aging-related muscle strength limitations. Implementation of 3D joint moment approach in the analysis of successful reactive-recovery was justified by the governing role of frontal plane joint moments.

5.1.6. *Hypothesis (f) - "The elderly will have lower JMG Rate (as of ankle, knee and hip) than the younger individuals during successful reactive-recovery."*

JMG Rate (joint moment generation rate), determined as the rate of JMP Magnitude (peak joint moment magnitude) to JMAP Time (joint moment activation to peak time), was used to characterize the speed of joint moment generation.

The proposed hypothesis was partially confirmed by the significant aging effect on JMG Rate found in several locations. In the ankle sagittal, knee sagittal and hip transverse planes, the younger group produced a significantly higher JMG Rate than their older counterparts during reactive-recovery. Such higher JMG Rate was mainly due to the significantly higher JMP Magnitude, given that no aging effect on JMAP Time was found. Previous studies considered the rate of lower extremity joint moment development as one important variable that govern our ability to recover balance (Chandler et al., 1990; Wojcik et al., 1999; Thelen, Wojcik, Schultz, Ashton-miller, & Alexander, 1997). Robinovitch et al (2002) studied the effect of speed of ankle torque development on balance recovery during lean-release experiments and concluded that human ability to recover balance following an unexpected perturbation was limited by the finite rates of torque generation. Although all the individuals in current study were able to achieve reactive-recovery, JMG Rates produced at ankle and knee sagittal plane by the younger group were still two times higher than those produced by the elderly. Such differences suggested that the age-related limitation in lower extremity muscle strength will expose these older individuals to higher risks of falls in future slip-induced accidents than their younger counterparts.

The finding of JMG Rate magnitudes supported those distinct balance recovery strategies summarized from JMP Ratio. The JMG Rates generated at the ankle sagittal and knee sagittal planes by the younger group were more than two times of those generated at ankle frontal and knee frontal planes, respectively. This result indicated the high dependence of sagittal plane joint moment by the younger group for balance recovery. However, the JMG Rates generated at the ankle sagittal and knee sagittal planes by the older group were roughly equal to those generated at ankle frontal and knee frontal planes.

As suggested in the balance recovery strategy adopted by the elderly, this result confirmed the considerable utilization of frontal joint moment to assist balance recovery for the elderly.

In summary, the decreased JMG Rate for the elder group found in ankle sagittal and knee sagittal planes indicated the age-related limitation of lower extremity muscle strength. Distinct balance recovery strategies adopted by young and old groups were also supported by the characteristic JMG Rates.

5.1.7. Hypothesis (g) - "The elderly will have lower JMP Power (as of ankle, knee and hip) than the younger individuals during successful reactive-recovery."

JMP Power (peak joint power magnitude) determined as peak joint power magnitude during stance phase, was used to enable further understanding of the role of lower extremity muscle groups from the energy perspective.

Gait condition effects on JMP Power were present in most of the locations. The peak joint powers generated by both age groups were found to be significantly higher during reactive-recovery than in normal walking. This finding indicated the higher needs for energy utilization in successful balance recovery. No significant gait effect was found in joint power generated at ankle sagittal and hip sagittal planes. By examining the descriptive summary of joint power (Table 4.9-1~9), such non significant differences were possibly due to the extremely high standard deviations. This highly variable joint power in ankle sagittal and hip sagittal planes can be plausibly explained by the individual-specific nature of reactive-recovery responses. Joint power analysis complements joint moment analysis by showing the summative mechanical power to/from the segments in the system by the muscle (Zajac, Neptune, & Kautz, 2002). Further complete joint power investigation together with joint work would provide a deep understanding of energy function of muscle groups surrounding lower extremity joints during reactive-recovery.

No aging effect on JMP Power was found in both normal gait and reactive-recovery. This finding suggested peak joint power as a common characteristic for all the individuals who were able to achieve successful reactive-recovery.

In summary, increased peak joint power during reactive-recovery was suggested as one common characteristic for all the individuals who were able to achieve successful reactive-recovery.

5.2. Conclusions and recommendations

Despite the individual-specific nature of reactive-recovery response, distinct age-related reactive joint moment development strategies were detected and summarized into three points.

First, during successful reactive-recovery, the younger and the elderly shared several common characteristics, which included joint moment activation time, joint moment activation to peak time, joint moment generation ratio and peak joint power. Increases in joint moment generation ratio during reactive-recovery suggested the possible benefit of lower extremity strength training in an effort to increase available muscle strength capability. Increases in peak joint power indicated the increased energy expenditure of the balance recovery response.

Second, elderly and younger participants adopted different joint moment generation strategies supported by the findings of peak joint moment magnitude, peak joint moment distribution ratio and joint moment generation rate. The younger individuals relied on the ankle and knee sagittal joint moment as the major actuator for balance recovery. However, due to the possible limitations of available strength capacity, the elderly highly depended upon ankle and knee frontal joint moment to achieve successful recovery.

Third, ankle joint was found to be the most critical joint regarding balance recovery, while hip joint assumed the major responsibility of balance maintenance of upper body, due to the findings associated with peak joint moment magnitude and peak joint moment distribution.

Moreover, implementation of 3D joint moment approach in the analysis of successful reactive-recovery was justified by the governing role of frontal plane joint moments. It can be reasonably concluded that future slip/fall research (including recovery and fall) requires 3D analysis if complete and accurate information is desired.

5.3. Assumptions

5.3.1. Identical data collection between two experiments

The data in current study was obtained from two experiments separated in 2002 and 2003. It was assumed these experimental data obtained were identical in nature because of the exactly same experiment protocol adopted. No year effect on walking velocity and step length during normal gait was found as shown in result chapter (Table 1). Such finding further confirmed the identical nature of experimental datasets used in current study.

5.3.2. Prior knowledge of existence of slippery surface

The current study applied a slippery surface to evoke unexpected slips. It was assumed that such an experimental setting would be able to simulate real-life slip/fall accidents. However, due to the protection of human participants, all the participants were informed of the existence of potential slippery surface. Such prior knowledge of slippery surface thus would lead to possible human gait adaptation (Cham et al., 2002a). For this reason, several approaches were utilized to minimize such gait adaptation. First, participants possessed no information about the exact location and time of the appearance of slippery surface. Second, each participant was allowed a normalization period of greater than 20 minutes to encourage natural gait, before introducing the slippery surface. Third, participants were further encouraged to walk normally by experiencing the fall-arresting harness. In addition, decision of implementing slippery surface was assisted by experimenter's subjective judgment based on observations of participants' walking style and walking speed. By applying the above approaches, it was reasonably assumed that the possible gait adaptation responding to prior knowledge of slippery surface was minimized.

5.4. Limitations

5.4.1. Possible contributions of arm movement and unperturbed limb

The focus of the current study was limited in the perturbed limb, which contacted the slippery surface, during reactive-recovery. However, a possible dynamic multi-limb coordinated strategy for dynamic stability might exist according to the findings by Marigold et al. in 2003. The importance of the unperturbed limb is first reflected in the findings that the critical time for stability recovery following the initiation of a slip is during the first double support phase of the gait cycle (You, Chou, Lin, & Su, 2001). Tang et al. (1998) observed that the unperturbed limb responds with rapid muscle activation and proposed possible inter-limb coordination in reactive-recovery. The unperturbed limb showed an extensor strategy supporting the lowering of the limb to touch the ground, to increase the base of support, and to increase stability (Marigold, Bethune, & Patla, 2003). The possible contribution of arm elevation following an unexpected slip was also highlighted (Marigold & Patla, 2002; Tang, Woollacott, & Chong, 1998). Following the unexpected slips, the arms are found to rapidly elevated forward and outward in an attempt to stabilize the backward shifted COM. Such arm elevation strategy has been documented in older adults (Tang & Woollacott, 1998) as well as young individuals (Marigold et al., 2002; You et al., 2001). Arm movements can also be used as a protective mechanism to avoid a potential injury from a fall (Maki & McIlroy, 1997; McIlroy & Maki, 1995). Therefore, future research on coordination of all the limbs during slip/fall accidents is expected to span our knowledge of successful reactive-recovery.

5.4.2. Adoption of exertion level at 120deg/s for strength measurement

Maximum isokinetic joint torque measurements were obtained at three exertion levels ($30^{\circ}\cdot s^{-1}$, $60^{\circ}\cdot s^{-1}$ and $120^{\circ}\cdot s^{-1}$). The computation of dependent variable JMG Ratio required the strength indicator. Choice of exertion level at $120^{\circ}\cdot s^{-1}$ was based on instant ankle sagittal angular velocity at heel contact. However, at the instant of peak joint moment (when the JMG Ratio was computed), the rotation speed of the corresponding joint was possible to deviate from $120^{\circ}\cdot s^{-1}$, which may lead to the deviation from the real strength estimates. Nevertheless, the current study showed significant positive correlations of strength measurement among the three exertion levels (Table 4.10-1). Thus, exertion level of $120^{\circ} \cdot s^{-1}$ as an indicator of strength capacity was sufficient for the purpose of current study. Certainly, in-depth strength estimates will be required in future slip/fall researches in which muscle strength investigation is a major objective.

5.4.3. Possible influence of fall-arresting harness

For the protection purpose, participants were required to wear fall-arresting harness to prevent their body hitting the ground, except foot segments. So far, no literature has documented the possible influence of using fall-arresting harness. But, influence of such harness on human gait, especially during slip and fall could be expected. Effort has been made to minimize the harness influence. Sufficient space on the harness cord was created to allow the participant enough room to accomplish balance recovery motions. Harness was designed to step in only after the participant's upright balance was totally lost. In summary, efforts have been made to minimize the possible influence of fall-arresting harness, though the complete elimination of harness can not be achieved at this stage due to ethics.

5.4.4. Hip marker configuration

There were two markers attached over left and right greater-trochanter landmarks. These two marker locations were used to estimate hip joints in current study. Such marker placements were limited by the utilization of fall-arresting harness. Future studies that are able to solve the harness limitation are expected to enable more sophisticated hip joint estimation.

5.4.5. Criteria for judging recovery and falls

There are no exact standards for judging whether the trials are reactive-recovery or falls. Therefore, generalization on findings from the current study has to be accepted with caution.

5.4.6. Limitations of inverse dynamics

Inverse dynamics is popular in estimating kinetic parameters among gait studies, and it is also the basis for the current study. However, several limitations existed in inverse dynamics might have an influence on the findings from the current study. First, rigid body segment is assumed. Second, soft tissue artifacts are neglected. Third, frictionless joint connection is assumed. Fourth, accuracy of joint estimation from external landmarks are still questionable.

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