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Doctoral Dissertation

Biomechanics Study on Age-Related Alterations of Walking Control

歩行制御の加齢変化における バイオメカニクス研究

July 2016

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Abstract

Walking ability of elderly people decreases with aging. Onset of the arthritic disease is associated with the mechanical stress applied to the joint during walking, and falling and fracture are caused by decreasing in balance while walking and walking control. By understanding the mechanisms of the age-related changes of walking, exercise regimen for the elderly can be developed. Therefore, it is possible for the elderly to prevent to become care state, and maintain their quality of life. Walking speed of the elderly is significantly slow due to a reduction in stride length associated with walking pattern decreased excursion of the hip and plantar flexion angle of the ankle. In kinetic analysis, elderly people show different distribution of joint moments and powers, with less joint moments and powers as compared with young people. Muscle activity of lower extremity during walking was generally higher for the elderly. These previous reports indicated that the elderly had different walking strategy from the young in addition to the reduction in physical function. In order to develop optimal exercise regimens for preventing reduction of ability to walk of the elderly, it is necessary to understand the age-related alterations of walking strategy. Moreover, the elderly has an inappropriate walking variability that indicates an inherent property of the motor system. Because maintained stability of the body during walking is performed by coordination between trunk and lower extremity, variability in the accelerations of the head, lumbar, and bilateral lower extremity may exhibit characteristics unique to each body part. In order to maximize the ability to determine the effects of age on walking, moreover, gender differences were necessary to be considered as well as age in these analyses.

In this dissertation, three experiments were performed in order to clarify the age-related alterations of walking control. Firstly, the stride-to-stride walking variability through simultaneous evaluation using accelerometer as a function of walking speed was examined (in Chapter 2). After that, the contribution of joint moment exerted by any joints of the lower extremity for the generation of the ground reaction force (in Chapter 3). Finally, the differences in the muscle control by central nerve system of the elderly and the effect on the support and propulsion during a gait cycle (in Chapter 4) were identified.

In the first research, the acceleration variability as a function of walking speed was focused. The aims were to reveal the stride-to-stride acceleration variability through simultaneous evaluation of the acceleration of the head, lumbar, and lower extremity. The effect of walking speed on this acceleration variability is different between the elderly and the young in each body part was also clarified. Twenty elderly people, 65 years old or older, and 20 young people, aged 20 to 29 years participated in this experiment. Eight accelerometers were used to measure the head, lumbar and bilateral lower extremity accelerations during walking. They were instructed to walk at five walking speeds prescribed by a metronome. Acceleration variability was assessed by cross-correlation coefficient normalized using Fisher's z transform. As a result, the effects of walking speed on the stride-to-stride acceleration variability of all body parts of the elderly and the young were same trend. Vertical acceleration might be strictly controlled so as to stabilize the walking behavior despite the effect of walking speed. On the other hand, the stride-to-stride acceleration variability of the lower extremity was affected when walking slowly both the elderly and the young in antro-posterior and medio-lateral directions. Especially in medio-lateral direction, there is laterality in the

effect of walking speed both the elderly and the young. Additionally, the elderly exhibited more repeatable acceleration patterns regardless of changing walking speed.

In the second research, the differences of the lower extremity joint moment and vertical and antero-posterior ground reaction forces during walking between the elderly and the young were examined. Furthermore, the differences in the relationships between the vertical and antero-posterior ground reaction forces and the lower extremity joint moment during walking between the elderly and the young were examined. Forty elderly people, 65 years old or older, and 40 young people, aged 20 to 29 years participated in this experiment. Joint moment of the lower extremity, and vertical and antero-posterior ground reaction forces during walking was obtained by a 3D motion analysis system and 8 force plates. A stepwise linear regression analysis determined the joint moments that predict the amplitude of vertical and antero-posterior ground reaction forces. As a result, the elderly had a significantly lower antero-posterior ground reaction force and the plantar flexion moment of the ankle than the young. On the other hand, the extension moment during the terminal stance phase of the knee was higher in the elderly participants. The young people both men and women mainly used knee joint moment for the control of both vertical and antero-posterior ground reaction forces. On the other hand, all of the hip, knee and ankle joint moments were related to these ground reaction forces in the elderly women. Therefore, the elderly and the young women used the lower limb joints differently for the control of vertical and antero-posterior ground reaction forces. In conclusion, the elderly had an age-related alteration in the strategy in order to generate support and propulsive force during walking.

In the third research, the differences of the amplitude, timing and combination of muscle activation during walking between the elderly and the young were examined.

Furthermore, the relationship between muscle modules and support and propulsive force during walking was also analyzed. Twenty elderly people, 65 years old or older, and 20 young people, aged 20 to 29 years participated in this experiment. Coordinates of anatomical landmarks and ground reaction force during walking were obtained by a 3D motion analysis system and 8 force plates. Muscle forces of lower extremity during walking were calculated using musculoskeletal simulation. Peak and average values of these muscles were compared between the elderly and the young, while muscle modules were obtained by analysis using non-negative matrix factorization. As a result, muscle activities of gluteus muscles of the elderly were excessive during stance phase. Moreover, in the elderly, there was a change in the coordination of muscles around the ankle, and muscles of the lower extremity exhibited co-contraction in late stance. Timing and shape of these modules were similar between elderly and young people in both men and women. In conclusion, intervention for muscle function during walking of the elderly should be performed not only for strengthening the weak muscle, but also for prevention some excessive muscle activities, and for changing the combination of muscle activity for each subtask of walking.

As conclusions of this dissertation, there were differences in walking control between the elderly and the young. The elderly had age-related alteration in control of lower extremity for support and propulsion during a walking cycle. Additionally, muscles of the lower extremity of the elderly exhibited co-contraction during walking, and the elderly performed more repeatable or stiff walking control as compared to the young regardless of changing walking speed. These study results revealed that control strategy of walking was changed with age, and the elderly had qualitative change in neuromuscular function of walking by central nerve system. It was pointed that,

rehabilitation therapists should pay attention to not only the muscle strength but also the control mechanism of walking, and it was necessary for ability to walk of the elderly to develop effective exercise regimen and treatment equipment for improvement in walking control by central nervous system.

Contents

Abstract i

Contents vi

List of Figures ix

List of Tables xi

CHAPTER 1 Introduction 1

1.1 Social background....1

1.2 Age-related alternations of musculoskeletal structure and physical function..3

1.3 Age-related alternations of walking....4

1.3.1 Spatiotemporal parameters....4

1.3.2 Walking variability....6

1.3.3 Kinematic parameters....8

1.3.4 Kinetic parameters....9

1.3.5 Muscle Activity....11

1.4 Questions not answered in previous studies....12

1.5 Objectives of this dissertation....15

1.6 Dissertation overview....16

CHAPTER 2 Analysis of acceleration variability through simultaneous evaluation of the head, lumbar and lower extremity as a function of walking speed 18

2.1 Purpose....18

2.2 Methods.....19

2.2.1 Subjects.....19

2.2.2 Experimental set up.....20

2.2.3 Procedure.....21

2.2.4 Data analysis.....22

2.2.5 Statistical analysis.....23

2.3 Results.....23

2.3.1 Walking speed effect.....25

2.3.2 Age effect.....27

2.4 Discussion.....34

2.5 Summary.....39

CHAPTER 3 Control of ground reaction force during walking by the hip, knee and ankle joint moments 41

3.1 Purpose.....41

3.2 Methods.....42

3.2.1 Subjects.....42

3.2.2 Experimental set up.....42

3.2.3 Data analysis.....44

3.2.4 Statistical analysis.....46

3.3 Results.....48

3.4 Discussion.....54

3.5 Summary.....60

CHAPTER 4 Muscle control of lower extremity during walking 61

4.1 Purpose.....61

4.2 Methods.....62

4.2.1 Subjects.....62

4.2.2 Experimental set up.....62

4.2.3 Gait evaluation.....63

4.2.4 Muscle force estimation.....64

4.2.5 Non-negative matrix factorization.....66

4.2.6 Statistical analysis.....68

4.3 Results.....69

4.4 Discussion.....76

4.5 Summary.....82

CHAPTER 5 Conclusion 84

5.1 Conclusion.....84

5.2 Future work.....87

Appendix A Calculation methods of joint moment 90

Appendix B Musculotendinous model 93

Appendix C Non-negative matrix factorization 96

Publications 98

References 100

Acknowledgments 114

List of Figures

Figure 1-1 Trends of aging in Japan……..……1

Figure 1-2 Causes the cornerstone care state of the elderly……2

Figure 1-3 Difference of walking pattern between the elderly and the young…..…...….8

Figure 1-4 A structure of muscle control……14

Figure 1-5 Concepts to assess the age-related alternations of walking control……..16

Figure 2-1 Wireless multi-function inertia sensor…….....19

Figure 2-2 Placement of accelerometers and acceleration sign convention.....20

Figure 2-3 Graphic representation of vertical acceleration at the level of the head, lumbar, and bilateral lower extremity.....23

Figure 2-4 Effect of walking speeds on the z-value in vertical (VT), antero-posterior (AP), and medio-lateral (ML) direction of the young.....32

Figure 2-5 Effect of walking speeds on the z-value in vertical (VT), antero-posterior (AP), and medio-lateral (ML) direction of the elderly.....33

Figure 3-1 Experimental set up of this study……...43

Figure 3-2 Flow of data analysis……44

Figure 3-3 Graphical representations of the vertical and antero-posterior ground reaction forces and the joint moment of the hip, knee, and ankle during a gait cycle……...47

Figure 3-4 Comparison of vertical and antero-posterior ground reaction forces (GRF) and lower extremity joint moments……..49

Figure 3-5 Models of the vertical and antero-posterior ground reaction forces by forward step-wise regression analysis of the elderly men and the young men……50

Figure 3-6 Models of the vertical and antero-posterior ground reaction forces by forward step-wise regression analysis of the elderly women and the young women ……51

Figure 3-7 The joint moments that can predict the amplitude of the vertical and antero-posterior ground reaction force in each group……..53

Figure 4-1 Musculoskeletal model……...……...63

Figure 4-2 Flow of muscle force estimation……64

Figure 4-3 Organization of the data matrix using NMF analysis……66

Figure 4-4 A schematic description of muscle control by modules……67

Figure 4-5 Muscle tension force during a gait cycle obtained by OpenSim.....72

Figure 4-6 Comparison of the peak values during a gait cycle, and average values during stance and swing phase of each muscle force between the elderly and young people both men and women……73

Figure 4-7 Variability accounted for (VAF) for determination of the number of synergies…….……..73

Figure 4-8 Muscle module weightings and activation timing profiles during walking derived from NMF analysis……..74

Figure 4-9 Vertical and antero-posterior ground reaction force profiles corresponding to each muscle module.....75

Figure A-1 A free body diagram of supporting leg during walking……...90

Figure B-1 Schema representation of the elements of a musculotendinous actuator as a single muscle in series with an elastic tendon.……..……...95

Figure C-1 Decomposition of a matrix by NMF……...96

List of Tables

Table 2-1 Characteristics of subjects…….25

Table 2-2 Z values of the young in vertical (VT), antero-posterior (AP), and medio -lateral (ML) directions for each speed conditions.....28

Table 2-3 Z values of the elderly in vertical (VT), antero-posterior (AP), and medio -lateral (ML) directions for each speed conditions.....30

Table 3-1 Characteristics of subjects…….48

Table 4-1 Characteristics of subjects…….68

Table 4-2 Peak values of joint moment.……69

Table 4-3 Similarities of activation pattern of muscle module between the elderly and the young……74

Table A-1 Body segment inertia parameters of the elderly and the young…….90

Table B-1 Muscle-tendon parameters for lower extremity muscles (Delp, 1990)……95

Chapter 1

Introduction

In this chapter, social backgrounds about the elderly, and mechanisms and age-related changes of walking are introduced.

1.1 Social background

In Japan, according to annual report on the aging society: 2014, the number of people aged 65 and older is the highest ever, at 31.9 million. Moreover, the percentage of the population aged 65 and older is 25.1% (**Figure 1-1**). Japan has entered super-aging

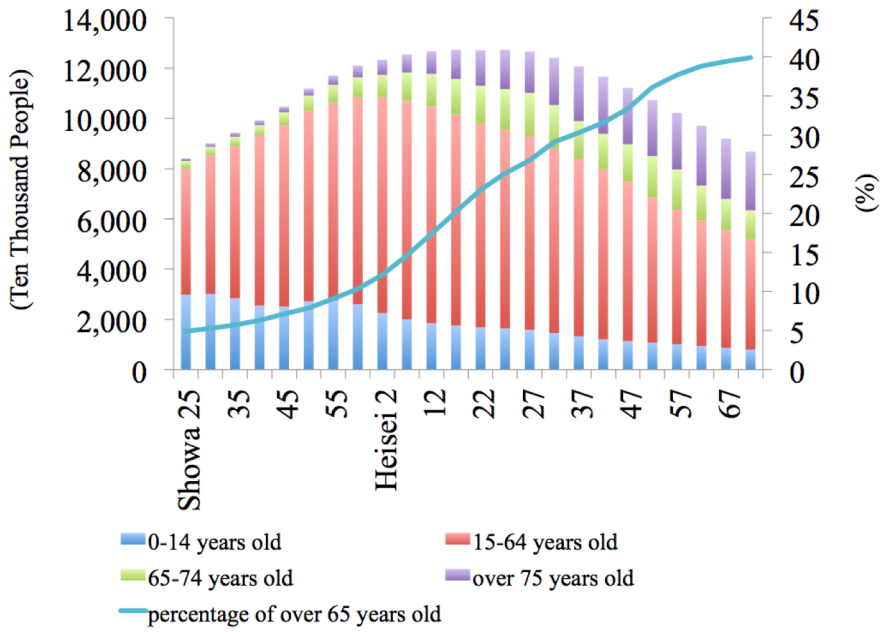


Figure 1-1 Trends of aging in Japan.

society from 2007. In the future, it is expected that the percentage of the elderly will continue to increase, as the total population decrease. To match the increase in the number of the elderly, elderly frail state is also increasing year by year. Thereby social security costs applied to the care of the elderly amounted to 107.495 trillion yen in 2011. It is able to improve physical and mental health of the elderly and reduce healthcare costs by prevention nursing care state of the elderly.

The elderly has several symptoms that influence their daily life (**Figure 1-2**). The most common cause is onset of disease such as stroke, cardiovascular disease and dementia. On the other hand, onset of arthritic disease is associated with mechanical stress applied to the joint during walking, and falling and fracture caused by a decrease

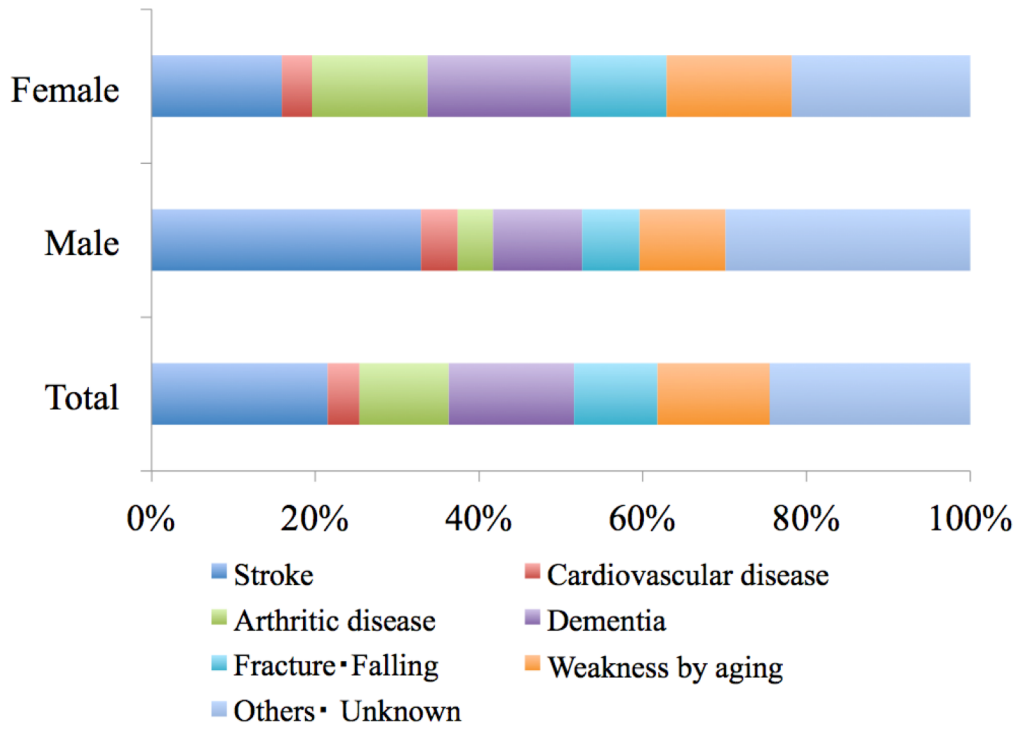


Figure 1-2 Causes the cornerstone care state of the elderly.

in balance while walking. Walking is frequently performed in daily life. Onset of these diseases of the elderly can be prevented by maintaining their ability to walk. Therefore, it is important for the development of exercise regimen of the elderly to understand the mechanism of the age-related changes of walking.

1.2 Age-related alternations of musculoskeletal structure and physical function

Elderly people often have spine deformity defined as thoracic and lumbar kyphosis, flat back and lumbar lordosis. Because malalignment of spine leads to abnormal postural control, patients with abnormal posture would be at increased risk for falling (Tinetti et al., 1988). Moreover, the elderly has significant losses in range of motion (Kuhlman, 1993). Flexibility declines 20-50% between the ages of 30 and 70 years (Chapman et al., 1972). The ability of muscle to produce an adequate amount of force is largely dependent on body structure and flexibility.

Muscle property of the elderly change with age. Janssen et al., (2000) reported that the loss of skeletal muscle mass with age was greater in the lower body in both men and women, and skeletal muscle expressed as a percent of body weight is reduced relatively early in life. In addition, there is no significant relationship between type Ⅰ fiber size and age, but there is a significant reduction in type Ⅱ fiber size with increasing age (Lexell et al., 1988). As a result, muscle strength declines with age in both men and women, especially in concentric peak torque (Porter, 1995). Muscle

strength is associated with walking ability (Steffen et al., 2002), disability and physical activity (Rantanen, et al., 1999), and risk of falling (Pijnappels, et al., 2008). Therefore, it is necessary for the elderly to prevent muscle weakness.

The aging process leads to change in the somatosensory system. Plantar tactile sensation is reduced in elderly people and associated with measures of balance and functional test performance (Menz, et al., 2005). In addition, there are morphological changes such as increasing capsular thickness and decreasing number of intrafusal muscle fibers (Swash, 1974). Thereby, static and dynamic muscle spindle sensitivity was declined with age. The elderly also has dysfunction of the visual systems (Saunders and Echt, 2007) and vestibular system (Fife and Baloh, 1993). These age-related changes in nerve system affect the control of joint motion, appropriate muscle activity, and balance control during walking.

1.3 Age-related alternations of walking

1.3.1 Spatiotemporal parameters

Many researchers reported about analyses of spatiotemporal parameters during walking of the elderly. The elderly aged 67 years old or older had significantly slower walking speed than young people for both their free and fast walking speed (Murray et al. 1969). Chung et al. (2010a) reported that the elderly had a 16% decrease in walking speed as compared with that of the young.

Stride length of the elderly aged 67 years old or older is also significantly less than that of the young for both walking speed (Murray et al. 1969). For comfortable walking speed, the mean stride length of subjects under 65 years old averaged 89% of their body height, whereas that of subjects over 65 years old averaged 79% of their body height. For fast speed walking, the mean stride length of subjects under 65 years old averaged 107% of their body height, whereas that of subjects over 65 years old averaged 90% of their body height. Therefore, the elderly cannot extend stride length at fast speed walking. On the other hand, there was no significant difference in cadence between the elderly and the young subjects (Winter et al., 1990). Slow walking speed of the elderly is not due to a decrease in cadence, but rather to a reduction in stride length.

Swing-to-stance time ratio of the elderly people aged 65 and older extend average stance duration of a walking cycle than the young in both comfortable and fast walking speed (Murray et al. 1969). Moreover, the time of double support phase increases in the elderly (Ferrandez et al., 1990). Therefore, the ratio of stance time is lower, and the time of double support phase is higher for the elderly.

1.3.2 Walking variability

Although walking is a periodic motion, there is variability in the kinematics and kinetics. In general, variability during walking has been believed to reflect the system stability. Therefore, walking variability should be maintained as small as possible (Masani et al., 2002), because excessive movement variability implies dynamic instability. Studies on walking variability have assessed the spatiotemporal characteristics (Jordan et al., 2007),

ground reaction forces (Masani et al., 2002), and acceleration patterns (Menz et al., 2003a). Increase in movement variability during walking is associated with the reduction in the coordination required for efficient walking control (Gabell and Nayak, 1984). Hausdorff et al. (2001) reported that stride time variability predicted falls in community-living older adults. Moreover, increase in stride-to-stride variability of the stride length, walking speed, and double support time were all independently associated with falling in the elderly subjects (Maki, 1997). Analysis of movement variability during walking may enhance the understanding of motor control and make it possible to predict one's stability of walk.

Movement variability during walking is affected by the walking speed. For expample, with respect to stride-to-stride fluctuation, Kang and Dingwell (2008) reported that variability of spatiotemporal characteristics and kinematic profiles were affected by the walking speed in both young and elderly people. They also reported that the effect of walking speed on variability as measured by frontal hip and knee motions, knee internal/external rotations, and trunk motions, were more pronounced at very high or very low speeds. Simirally, Kavanagh (2009) reported that medio-lateral and vertical trunk accelerations have less signal regularity and repeatability during slow compared to preferred speed walking. Walking variability must be properly controlled irrespective of the change in walking speed in order to minimize perturbations to the body during walking. It is, therefore, important to clarify the relationship between the acceleration variability and walking speed.

Accelerometer measures the accelerations and decelerations of body segments. Because accelerometer is low cost, compact in size, and less restrictive than motion analysis system in a laboratory, analysis using accelerometers can be performed in living environment as well as in clinical settings without hindering normal movements. Evaluation of the variability during walking using accelerometer with such participants as community-living older adults (Hausdorff et al., 2001), stroke patients (Mizuike et al., 2009), adults with symptomatic knee osteoarthritis (Tochigi et al., 2009), and neuropathic patients (Menz et al., 2004) have been performed. Evaluation of the variability of walking using an accelerometer is clinically useful.

Age-related alteration of acceleration variability is evaluated by root mean square, correlation and entropy analysis. The smoothness of acceleration patterns is very similar, although the magnitude of accelerations at the head and pelvis are smaller in the elderly (Menz et al., 2003b). Frail older adults have lower medio-lateral inter stride trunk variability, which may play a role in locomotors control in older individuals than fit older adults, but higher variability in the direction of propulsion (Moe-Nilssen and Helbostad, 2005). Walking variability associated with increasing risk of falling of the elderly. Moreover, walking variability relate to the attention and cognitive function of the elderly. Asai et al. (2013) reported that whole trunk control in the medio-lateral direction might be strongly associated with attention. There is an association between executive tasks and trunk acceleration pattern in the elderly with dementia (Ijmker and Lamoth, 2012). These findings may indicate that it can be easily performed evaluation of the walking stability of the elderly by evaluation of acceleration variability.

1.3.3 Kinematic parameters

Changes in walking pattern with aging are features of walking in the elderly (**Figure 1-3**). Vertical pathway of body center of mass is decreased, and lateral pathway of body center of mass is increased (Murray et al. 1969). With regard to joint angle profiles, in hip joint, the elderly has less total excursion (Murray et al. 1969; Toda et al., 2013), maximum extension angle in late stance (Kerrigan et al., 1998a), and greater flexion angle variation during late stance to toe-off (Toda et al., 2013). In knee joint, the elderly has less maximum flexion angle during swing phase (Murray et al. 1969), and greater flexion angle variation during late stance to toe-off (Toda et al., 2013). In ankle joint, the elderly has less maximum plantar flexion angle at toe-off (Murray et al. 1969), plantar flexion angle variation during stance phase (Toda et al., 2013), and length of

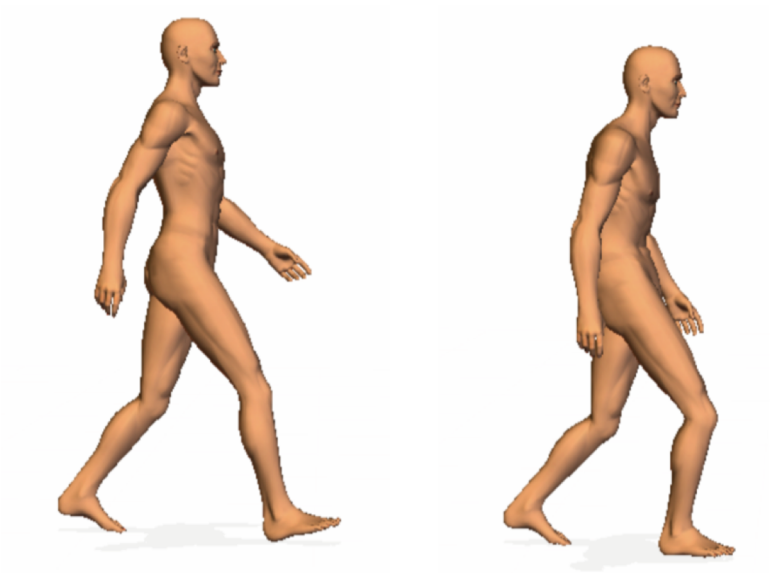


Figure 1-3 Difference of walking pattern between the elderly and the young (Murray et al., 1969).

elevation of toe (Kaneko et al., 1991). In summary, features of walking pattern of the elderly are flexed position of the hip and knee, and less hip extension and ankle plantar flexion angle.

1.3.4 Kinetic parameters

Ground reaction force is an external force involved in walking, which affects the acceleration of body center of mass (Winter 2009). Vertical and antero-posterior ground reaction forces are often measured in gait analysis, and regarded as a representative measure of walking (Jacobs, et al., 1972). The two peaks of vertical ground reaction force during early stance and late stance reflect the support of body center of mass. The minimum value at mid-stance indicates a reduction of the vertical force applied on the ground, and thus the reaction force from the ground to the body also decreases (Liu et al., 2006). In the first half of stance, posterior component of ground reaction force decelerates the body center of mass as a braking force. In the second half of stance, anterior component of ground reaction force accelerates the body center of mass as a propulsive force.

It is also known that ground reaction force during walking is affected by aging. Elderly people have low first peak and second peak of the vertical ground reaction force, braking, and propulsive forces, and high minimum value at mid-stance as compared with young people (Yamada and Maie, 1988). The profiles of the vertical and antero-posterior ground reaction force are correlated with many parameters of gait and functional performance. M-shaped vertical ground reaction force of elderly women was

associated with good results of Timed Up and Go test and Functional Reach test (Takahashi et al., 2004). In addition, Nilsson and Thorstensson (1989) reported that increases in the antero-posterior ground reaction forces were accompanied by increased walking speed. Yamada and Maie (1988) reported that the antero-posterior ground reaction forces are diminished in elderly persons, as compared with young people. Thus, antero-posterior ground reaction forces can be considered to be factors affecting walking speed in elderly people. Consequently, the profiles of the ground reaction force provide correlative information about the ability to walk of the elderly people.

By analysis of joint moment profiles, it is possible to consider the difference in force exerted by the joints. Hip and knee joint extension moments, knee power generation and absorption, and ankle power generation of the elderly are significantly lower than those of younger people (Kerrigan et al., 1998a; Winter, 1990). Elderly people who may be at risk of falls have a significantly large hip flexion moment in stance phase, and less hip extension moment, knee flexion moment in pre-swing, and knee power absorption in pre-swing (Kerrigan et al., 2000). A decline in plantar flexor moment and power of the elderly relates to slower walking speed (Judge et al., 1996, Winter, 1980). Moreover, the elderly people show different distribution of joint moments and powers as compared to young people. Support during stance phase is achieved by net extensor moments at the ankle, knee, and hip joint (Winter, 1980). DeVita and Hortobagyi (2000) reported that the elderly people walked at the same speed as the young people by producing the same support torque, but they produced this total output with a larger contribution from hip extension moment and smaller

contributions from knee extension and ankle plantar flexion moments. In addition, the elderly produced more work with their hip extensors and less with their knee extensors and ankle plantar flexors as compared with the young people. In the elderly, the reduction of the hip and knee moments contribution to hip linear power is significant as compared with the young at their normal gait speed. However, when asked to walk fast, the elderly significantly increase knee and hip moment contributions to hip linear power (Riley et al., 2001). Therefore, joint kinetics is age-related changes in both qualitative and quantitative aspects.

1.3.5 Muscle activity

Muscle activity analysis is generally performed using electromyography (EMG). Muscle activity during walking of the elderly becomes excessive. Finley et al. (1969) reported that muscle activities of biceps femoris, tibialis anterior, peroneus longus and gastrocnemius were higher for the elderly women than for those of the young at comfortable walking. Yanagawa et al. (2002) reported that integrated EMG of rectus femoris, vastus medialis, biceps femoris and tibialis anterior of the elderly men were significantly higher than those of the young at comfortable and similar walking speeds. In addition, the activity duration of their rectus femoris, vastus medialis and tibialis anterior were prolonged than those of young subjects. Schmiz et al., (2009) analyzed the muscle activation of the lower extremity during walking when walking speed was changed. As a result, the elderly exhibited larger activation of the tibialis anterior and soleus during mid-stance at the slow, preferred and fast walking speeds, and larger

activation of the vastus lateralis and medial hamstrings during loading response to mid-stance at the fast walking speed. They concluded that the elderly demonstrated significantly greater co-activation across the ankle and knee, and the age-related changes in neuromuscular activity reflect a strategy of stiffening the limb during single limb support. In addition, because soleus muscle activation for push off of the elderly is not able to increase at faster walking speed, muscle weakness of soleus likely contributed to reduced push off power. Muscle activity during walking of the elderly becomes excessive and greater co-activation.

1.4 Questions not answered in previous studies

There are some questions about the age-related changes of walking control in previous studies. Firstly, previous studies have been made on acceleration variability of the head, lumbar, and/or pelvis independently (Menz et al., 2003a, 2003b; Kavanagh, 2009). However, it is assumed that the acceleration variability of the head and trunk are affected by the control strategy of the lower extremity in a concerted manner. Rotational moments delivered from lower extremity to the trunk is controlled to be small, and trunk is adjusted in an upright position so that the gravity term become nearly zero (Winter, 1995). Furthermore, the acceleration from pelvis to head is absorbed by the spinal column (Kavanagh, 2004). Thus, the acceleration variability of the head, lumbar, and the lower extremity may exhibit characteristics unique to each body part. By simultaneous evaluation of the acceleration variability of the lower extremity in addition

to the head and pelvis, it is able to understand the difference in adaptation strategy to changes in walking speed between the head, lumbar, and lower extremities. Therefore, in Chapter 2, the purpose is to examine the stride-to-stride acceleration variability through simultaneous evaluation of the acceleration of head, lumbar, and lower extremity, and to clarify if the effect of walking speed on these acceleration variability is different between the body parts in the elderly and the young.

Secondly, vertical and antero-posterior components of ground reaction force are important for generation of support and propulsive force during walking, and forces that antagonize the forces that are exerted by the flexion-extension moment of the lower extremity. On the other hand, the elderly people show different distributions of joint moments and powers compared to the young people (DeVita and Hortobagyi, 2000). If elderly people exhibit an age-related alteration in moment of the lower extremity, changes in joint moment may affect the magnitude of the ground reaction force. By analyzing the relationships between vertical and antero-posterior ground reaction force and the lower extremity joint moments, it would be possible to clarify the difference in the strategies that generate ground reaction force of elderly and young people. Therefore, in Chapter 3, the purpose is to examine the relationships between the vertical and antero-posterior ground reaction forces and the hip, knee, and ankle joint moments during walking by the elderly and young people.

Thirdly, joint moment of lower extremity is generated by controlling muscle activity. Previous study reported that muscle activation patterns during walking of the elderly change with age without quantitative analysis (Finley et al., 1969). Muscle activity during motion is adjusted to appropriate amount, timing and combination of muscle activation by the networks in central nervous system (**Figure 1-4**). Outcome of the training is dependent on the training task and the context itself (Sale and MacDougall, 1981), although strength training is recommended to prevent the decline of walking ability of the elderly. In order to develop optimal exercise regimens to prevent reduction of walking ability of the elderly, it is necessary to understand the differences of the amplitude, timing and combination of muscle activation during walking. Therefore, in Chapter 4, the purpose is to examine the age-related differences of the muscle control during walking.

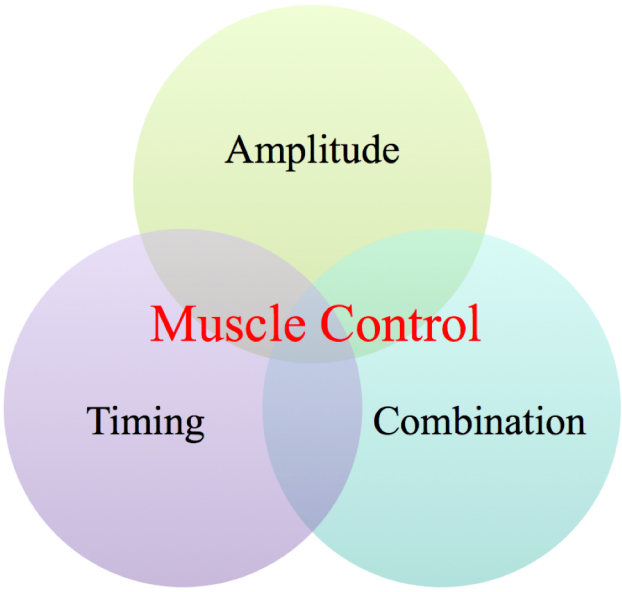


Figure 1-4 A structure of muscle control.

Finally, previous studies reported that there were significant gender differences associated with body structures (Cho et al., 2004; Iglic et al., 2001), walking parameters (Oberg et al., 1993; Kerrigan et al., 1998b; Chung et al., 2010b), and onset of disease (Yoshimura et al., 2009 and 2013). However, many previous studies examined the differences in walking parameters between the elderly and the young without consideration of gender factor. Therefore, in Chapter 2, 3 and 4, analyses are performed in consideration of gender differences as well as age.

1.5 Objectives of this dissertation

Figure 1-5 shows concept to assess the age-related alternation of walking control. Central nerve system generates a force and inhibits variability by controlling the muscle of the lower extremity. In order to clarify the age-related alternation of walking control, this dissertation has three purposes as follows:

1) To examine the walking variability through simultaneous evaluation of the head, lumbar, and lower extremity using accelerometer as a function of walking speed (Chapter 2).
2) To investigate the relationship between the joint moment, which is exerted by any joints of the lower extremity and the generation of ground reaction force (Chapter 3).
3) To clarify the difference in the muscle control between the elderly and the young, and the effect on the support and propulsion of walking (Chapter 4).

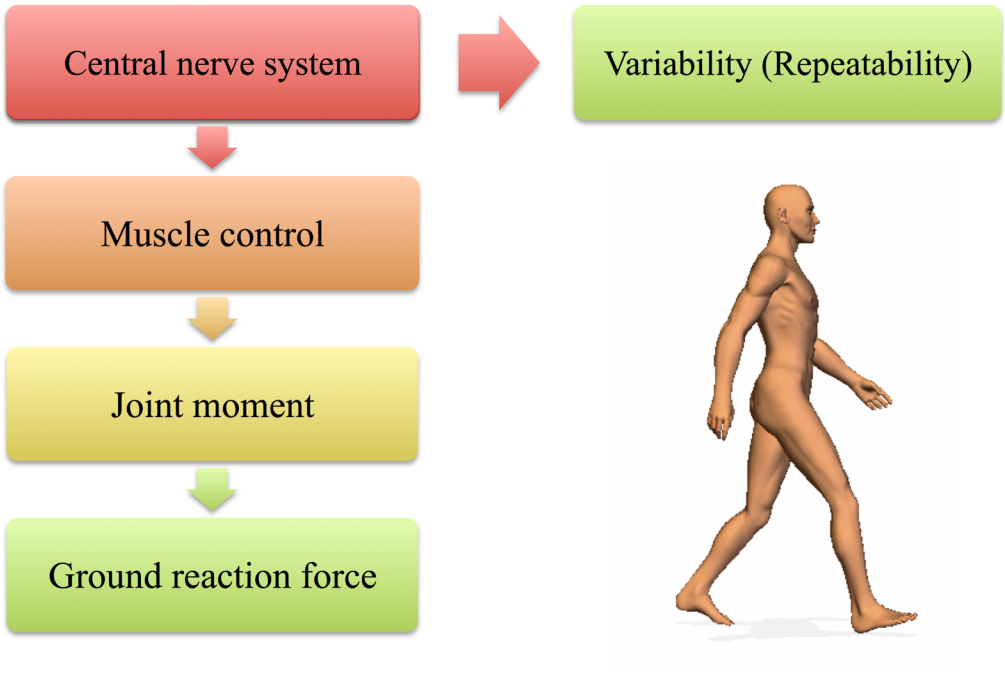


Figure 1-5 Concepts to assess the age-related alternations of walking control.

1.6 Dissertation overview

This dissertation is organized as follows: Chapter 1 presents the research area of the age-related alternations of walking and looks at previous research in this area. Chapter 2 investigates the variability in walking through simultaneous evaluation of the acceleration of whole body. The differences in the acceleration variability of the head, lumbar, and lower extremity as a function of walking speeds between the elderly and the young are clarified. Chapter 3 investigates the age-related difference in the walking strategy between the elderly and the young. The differences in the relationships between the vertical and antero-posterior ground reaction force and the hip, knee, and ankle joint

moments during walking between the elderly and the young are clarified. Chapter 4 investigates muscle control during walking in the elderly and the young by using musculoskeletal simulation. The differences in the amplitude of muscle force and the muscle modules of the lower extremity during walking between the elderly and the young are clarified. Finally, Chapter 5 provides conclusions on age-related alternation of walking control.

Chapter 2

Analysis of acceleration variability thorough simultaneous evaluation of head, lumbar and lower extremity as a function of walking speed

This chapter describes the walking variability of whole body of the elderly and the young using accelerometer in several walking speed. The aims are to find out the differences in the acceleration variability through simultaneous evaluation of the head, lumbar, and lower extremity as a function of walking speed between the elderly and the young. Twenty elderly people, 65 years old or older, and 20 young people, aged 20 to 29 years participated in the experiment.

2.1 Purpose

The purpose of this study is to investigate the stride-to-stride acceleration variability through simultaneous evaluation of the acceleration of head, lumbar, and lower extremity, and to clarify whether the effect of walking speed on the acceleration variability is different between the elderly and the young. Moreover, the differences in

the stride-to-stride acceleration variability of each body part between the elderly and the young are also evaluated. It is hypothesized that the variability of accelerations in the lower extremities of the elderly is affected by the walking speed, although those of the head and lumbar are well controlled irrespective of changes in the walking speed. It is also hypothesized that the variability of accelerations of the lower extremities of the elderly was larger than the young.

2.2 Methods

2.2.1 Subjects

Twenty community-dwelling elderly people, 65 years old or older and 20 young people, aged 20 to 29 years participated in this study. There were 10 men and 10 women in each group. The subjects were all right-handed. The exclusion criteria included neurologic disorders, osteoarthritis, rheumatic arthritis, joint pain affecting walking, and a history of surgery in the lower extremities or spine. All procedures were approved by the Ethics

37m (W) × 46mm (H) ×12mm (D), 22g

Figure 2-1 Wireless multi-function inertia sensor.

Committee at Kurashiki Heisei Hospital, and all participants provided their written informed consent prior to enrollment.

2.2.2. Experimental set up

Eight wireless multi-function inertia sensors (TSND121; 37-mm width, 46-mm height, 12-mm depth; weight, 22g; ATR-Promotions Co., Ltd., Kyoto, Japan (**Figure 2-1**)) were used to measure head, lumbar, and lower extremity accelerations along three axes (vertical (VT), medio-lateral (ML), and anterio-posterior (AP)) at a sampling frequency of 100 Hz and a measurement range of ± 8G. Placement of accelerometers and acceleration sign convention are shown in **Figure 2-2**. The wireless sensors were placed at external occipital protuberance (head), L3 processus spinosus (lumbar), bilateral

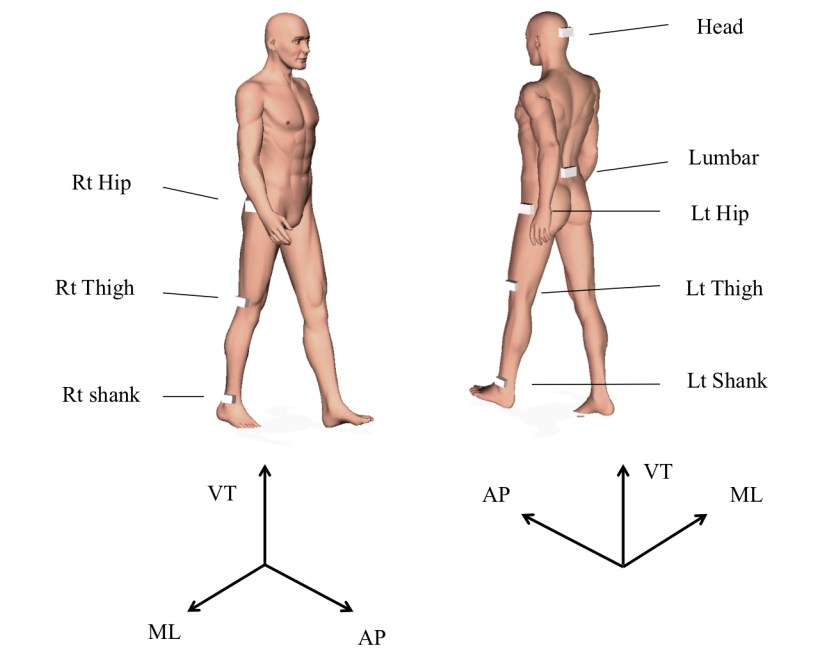


Figure 2-2 Placement of accelerometers and acceleration sign convention.

mid-point between the crest of the ilium and the great trochanter (hip), bilateral distal outer thigh (thigh), and bilateral distal outer shank (shank). Data were stored in the internal memory. After the measurement was completed, these data were transferred to a personal computer via Bluetooth.

2.2.3. Procedure

In the walking trials, all participants were instructed to walk with bare feet, to the end of a 15-m long walkway, and turn around, and walk back. Therefore, a total distance of 30 m was covered. To ensure that the steps and walking motion were stable, the first and the last two-to-three steps were excluded from the analysis, because two or three steps are necessary in the transition between gait initiation and steady-state walk of healthy people (Mann et al., 1979). Before collecting any data from the participants, cadence and 10-m walking time during steady-state walking were measured at their preferred speed. Ten-meter walking time was recorded in seconds using a digital stopwatch. The number of steps to walk 10m was used to calculate the average cadence (steps/min). Referring to comfortable cadence of walking (100%), the walking speed was prescribed by a metronome at 80, 90, 100, 110, and 120% for the very low, low, preferred, fast, and very fast walking speed conditions, respectively, based on a preceding study (Jordan et al., 2007). Walking speeds were prescribed at random.

2.2.4. Data analysis

To provide an indication of the variability of acceleration patterns between strides, acceleration signals were divided into individual strides by analyzing the vertical component. The identification of particular points that correspond to heel contacts in the acceleration was performed using the method reported by Auvinet et al. (**Figure 2-3**).

Ten acceleration data during straight line walking were extracted after initial heel contact was determined in each walking trial. The acceleration waveform of each walk cycle was treated as an independent time series data in this study, and the stride-to-stride acceleration variability was analyzed using cross-correlation analysis. When time series of a walking cycle x_i and time series of other walking cycle y_i are considered, cross-correlation C_{xy} is given by

$$C_{xy}(r) = \frac{\sum_{i=1}^{N-k}(x_i - \bar{x})(y_{i+k} - \bar{y})}{\sqrt{\sum_{i=1}^{N-k}(x_i - \bar{x})^2}\sqrt{\sum_{i=1}^{N-k}(y_{i+k} - \bar{y})^2}} \tag{2.1}$$

where N is series length, $\bar{x}$ and $\bar{y}$ are sample means, k is time lag. Cross-correlation can be useful for investigating the similarities of two-time series data sets (Nelson-Wong E et al., 2009). For the cross-correlation analysis in VT, AP and ML directions, each stride was time normalized (0–100%), and forty-five combinations of the 10 strides were obtained (${}_{10}C_2 = 45$). Cross-correlation coefficients were calculated in each pair. These coefficients were normalized using Fisher's r to z-transform (z-value), which yields variates that are approximately normally distributed, avoiding floor and ceiling effects (Press, et al., 1992):

$$Z\ (r) = \frac{1}{2} log_e \frac{1+r}{1-r} \tag{2.2}$$

Z-values of 45 combinations were averaged. A high z-value indicates low variability, since there is a strong correlation between each stride. All calculations were performed by Matlab R2013a (The MathWorks, Inc., Natick, MA, USA).

2.2.5. Statistical analyses

A four-way analysis of variance with walking speed, directions, age and gender was used to compare z-values for each body part. Differences between walking speeds were determined by Tukey's honest significant difference test. Statistical significance was set at $p < 0.05$. All data were processed by SPSS 17.0 statistical software (SPSS Japan Inc., Tokyo, Japan).

2.3. Results

The baseline demographic characteristics of the study subjects are presented in **Table 2-1**. The elderly participants had significantly shorter body height and higher BMI than the young study participants, but there was no significant difference in their body masses. The results of comparisons of z-values with each speed conditions are shown in **Table 2-2** and **Table 2-3**, and the results of this study are shown in **Figure 2-4** and **Figure 2-5.**

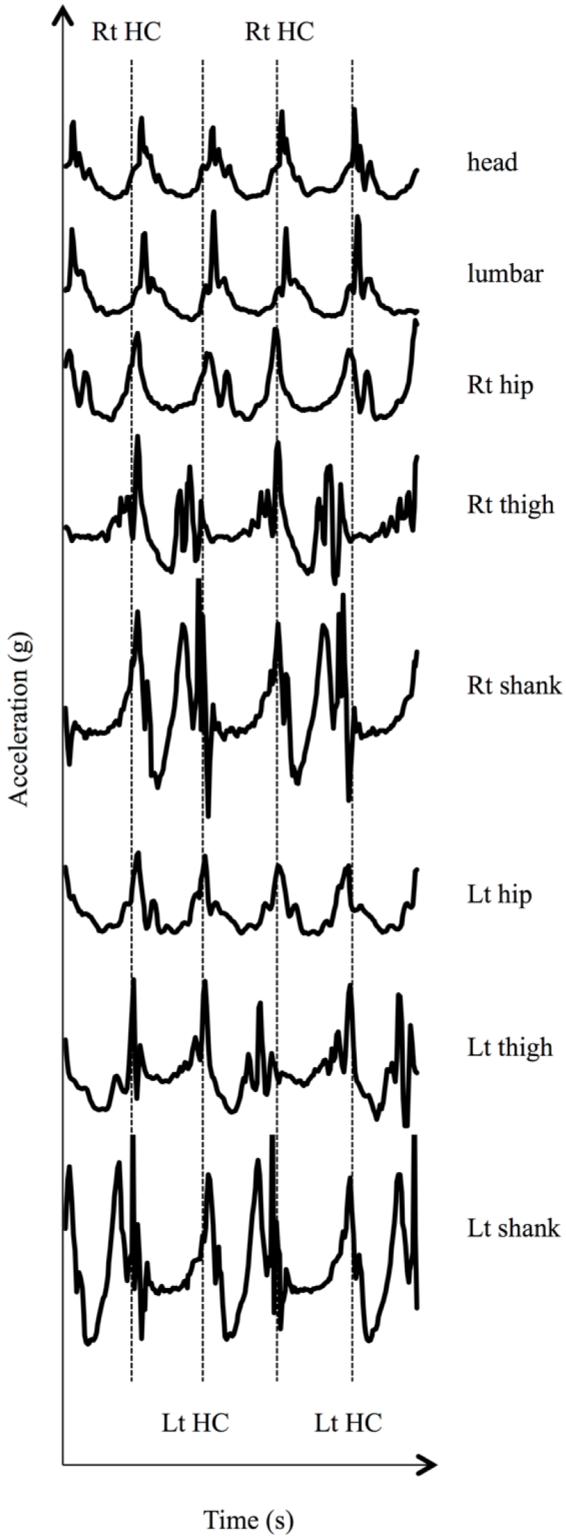


Figure 2-3 Graphic representation of vertical acceleration at the level of the head, lumbar, and bilateral lower extremity. The dashed line corresponds to right (Rt) and left (Lt) heel contact (HC).

Table 2-1 Characteristic of subjects.

Variable	Elderly		Young	
	Men (n = 10)	Women (n = 10)	Men (n = 10)	Women (n = 10)
Age (years)	78.8 (7.8)*	83.5 (5.3)*	25.4 (2.5)	26.4 (2.0)
Height (m)	1.57 (0.05)*	1.45 (0.05)*	1.67 (0.05)	1.59 (0.06)
Weight (kg)	58.9 (9.5)	48.1 (9.8)	56.8 (5.5)	50.3 (8.6)
BMI †(kg/m^2)	23.7 (3.8)*	22.7 (4.2)*	20.2 (2.1)	19.6 (2.7)
Walking speed (m/s)	0.94 (0.15)*	0.95 (0.12)*	1.28 (0.14)	1.27 (0.07)

Value: mean (Standard deviation: SD)

* : Significant difference from the young at $p < 0.05$

† : BMI: Body Mass Index

‡ : % Body Height

2.3.1 Walking speed effect

There were no significant interactions between speed, age and gender factors in all body parts. Therefore, the young and the elderly showed the same trend. In the head, z-value of very slow speed was significantly larger than the value for very fast speed in all directions.

In the lumbar, z-value of very slow speed was significantly larger than the value for the value for slow, fast and very fast speeds in vertical direction. In AP direction, there was not significant effect of walking speed factor. Z-value of slow speed was significantly lower than the value for preferred and very fast speeds in ML direction.

In the right hip, z-value of very slow speed was significantly larger than the value for slow and very fast speeds, and z-value of slow condition was significantly lower than the value for very slow and preferred in all directions.

In the right thigh, z-value of very slow speed was significantly larger than the value for the value for slow, fast and very fast speeds in vertical direction. In AP direction, z-value of very slow speed was significantly larger than the value for the value for slow and fast speeds, and z-value of slow speed was significantly lower than the values for other walking speeds. Z-value of slow condition was significantly lower than the values for preferred, fast and very fast speed speeds in ML direction.

In the right shank, z-value of slow speed was significantly lower than the values for very low, preferred and fast speed speeds, and z-value of preferred speed was significantly larger than the values for very fast speed in vertical direction. Z-values of very low and low speeds were significantly lower than the values for preferred, fast and very fast speeds in AP direction. Z-value of low condition was significantly lower than the values for preferred, fast and very fast speed speeds in ML direction.

In the left hip, z-value of very slow speed was significantly larger than the value for slow and very fast speeds in vertical direction. In AP direction z-value of low speed was significantly lower than the value for preferred, fast and very fast speeds. There was not significant effect of walking speed factor in ML direction.

In the left thigh, z-value of very slow speed was significantly larger than the value for slow and very fast speeds in vertical direction. Z-values of very low and low speeds were significantly lower than the values for preferred, fast and very fast speeds in AP direction. There was not significant effect of walking speed factor in ML direction.

In the left shank, z-value of low speed was significantly lower than the values for very low and preferred speed speeds in vertical direction. Z-values of very low and low speeds were significantly lower than the values for preferred, fast and very fast speed speeds in AP direction. There was not significant effect of walking speed factor in ML direction.

2.3.2 Age effect

In the head, Z-values of the elderly in vertical and ML directions were significantly larger than the values for the young, although z-values of the elderly in AP direction were significantly lower than the values for the young. In the lumbar, right and left shank, z-values of the elderly in all directions were significantly larger than the values for the young. In the right and left hip, z-values of the elderly in vertical and AP directions were significantly larger than the values for the young. In the right thigh, z-values of the elderly in vertical direction were significantly larger than the values for the young, although z-values of the elderly in ML direction were significantly lower than the values for the young. In the left thigh, z-values of the elderly in vertical direction were significantly larger than the values for the young.

In the head, lumbar, right hip and left thigh, z-values of the elderly women were significantly larger than the values for the young women.

Table 2-2 Z values of the young in vertical (VT), antero-posterior (AP), and medio-lateral (ML) directions for each speed conditions.

		very low		low		preferred		fast		very fast	
		Men	Women	Men	Women	Men	Women	Men	Women	Men	Women
Head	VT	2.70 (0.21)$^{\|}$	2.74 (0.18)$^{\|}$	2.54 (0.22)	2.45 (0.19)	2.48 (0.27)	2.47 (0.31)	2.29 (0.34)	2.27 (0.22)	2.32 (0.31)*	2.23 (0.13)*
	AP	2.12 (0.40)$^{\|}$	1.43 (0.74)$^{\|}$	2.13 (0.33)	1.29 (0.55)	2.11 (0.39)	1.47 (0.68)	2.00 (0.41)	1.45 (0.69)	2.07 (0.41)*	1.25 (0.61)*
	ML	1.36 (0.25)$^{\|}$	1.41 (0.45)$^{\|}$	1.17 (0.30)	1.07 (0.30)	1.27 (0.22)	1.25 (0.36)	1.24 (0.22)	1.26 (0.35)	1.31 (0.26)*	1.28 (0.41)*
Lumbar	VT	2.76 (0.26)$^{\dagger,\S,\|}$	2.78 (0.19)$^{\dagger,\S,\|}$	2.46 (0.23)*	2.41 (0.21)*	2.49 (0.12)	2.58 (0.30)	0.45 (0.25)*	2.34 (0.26)*	2.32 (0.24)*	2.36 (0.16)*
	AP	1.50 (0.28)	1.69 (0.23)	1.29 (0.26)	1.33 (0.24)	1.37 (0.29)	1.67 (0.38)	1.42 (0.30)	1.47 (0.20)	1.27 (0.23)	1.55 (0.29)
	ML	1.02 (0.20)	1.17 (0.22)	0.89 (0.18)$^{\ddagger,\|}$	0.96 (0.18)$^{\ddagger,\|}$	0.98 (0.16)†	1.27 (0.30)†	1.11 (0.31)	1.12 (0.22)	1.01 (0.19)†	1.21 (0.22)†
Rt Hip	VT	2.52 (0.27)$^{\dagger,\|}$	2.52 (0.21)$^{\dagger,\|}$	2.33 (0.23)*,‡	2.19 (0.18)*,‡	2.30 (0.19)†	2.34 (0.21)†	2.21 (0.29)	2.23 (0.33)	2.08 (0.21)*	2.24 (0.21)*
	AP	1.27 (0.32)$^{\dagger,\|}$	1.36 (0.25)$^{\dagger,\|}$	1.06 (0.42)*,‡	1.06 (0.25)*,‡	1.29 (0.22)†	1.40 (0.26)†	1.25 (0.38)	1.29 (0.34)	1.20 (0.22)*	1.37 (0.24)*
	ML	1.81 (0.41)$^{\dagger,\|}$	2.24 (0.46)$^{\dagger,\|}$	1.58 (0.50)*,‡	1.95 (0.43)*,‡	1.87 (0.75)†	2.08 (0.40)†	1.78 (0.71)	1.95 (0.39)	1.79 (0.96)*	1.93 (0.28)*
Rt Thigh	VT	2.28 (0.19)$^{\dagger,\S,\|}$	2.55 (0.31)$^{\dagger,\S,\|}$	2.11 (0.20)*	2.21 (0.28)*	2.14 (0.20)	2.28 (0.25)	2.15 (0.30)*	2.23 (0.23)*	1.96 (0.22)*	2.25 (0.24)*
	AP	1.21 (0.27)†,§	1.45 0.21)†,§	1.08 (0.26)$^{*,\ddagger,\S,\|}$	1.15 (0.30)$^{*,\ddagger,\S,\|}$	1.31 (0.25)*	1.40 (0.31)*	1.48 (0.31)*,†	1.47 (0.27)*,†	1.36 (0.28)†	1.53 (0.19)†
	ML	1.16 (0.18)	1.59 (0.41)	1.07 (0.20)$^{\ddagger,\S,\|}$	1.35 (0.40)$^{\ddagger,\S,\|}$	1.18 (0.27)†	1.47 (0.32)†	1.37 (0.26)†	1.49 (0.27)†	1.24 (0.20)†	1.58(0.24)†
Rt Shank	VT	2.26 (0.28)†	2.29 (0.25)†	1.99 (0.17)*,‡,§	2.04 (0.26)*,‡,§	2.25 (0.16)$^{\dagger,\|}$	2.33 (0.23)$^{\dagger,\|}$	2.11 (0.17)	2.25 (0.21)	2.02 (0.20)*,‡	2.12 (0.25)*,‡
	AP	1.25 (0.23)$^{\ddagger,\S,\|}$	1.30 (0.18)$^{\ddagger,\S,\|}$	1.15 (0.22)$^{\ddagger,\S,\|}$	1.12 (0.23)$^{\ddagger,\S,\|}$	1.51 (0.21)*,†	1.57 (0.21)*,†	1.45 (0.26)*,†	1.60 (0.15)*,†	1.55 (0.25)*,†	1.51 (0.28)*,†
	ML	1.16 (0.18)	1.11 (0.13)	1.08 (0.15)$^{\ddagger,\S,\|}$	0.98 (0.14)$^{\ddagger,\S,\|}$	1.32 (0.20)†	1.23 (0.17)†	1.21 (0.24)†	1.24 (0.15)†	1.26 (0.28)†	1.27 (0.16)†

(continued on next page)

Values: Mean (Standard Deviation)

*: Significant difference from very slow condition at $p < 0.05$, †: Significant difference from slow condition at $p < 0.05$, ‡: Significant difference from preferred condition at $p < 0.05$, §: Significant difference from fast condition at $p < 0.05$, ‖: Significant difference from very fast condition at $p < 0.05$.

Table 2-2 (CONT'D) Z values of the young in vertical (VT), antero-posterior (AP), and medio-lateral (ML) directions for each speed conditions.

		very low		low		preferred		fast		very fast	
		Men	Women	Men	Women	Men	Women	Men	Women	Men	Women
Lt Hip	VT	2.45 (0.15)$^{\dagger,\|}$	2.51 (0.22)$^{\dagger,\|}$	2.25 (0.26)*	2.20 (0.20)*	2.33 (0.25)	2.45 (0.41)	2.34 (0.24)	2.39 (0.26)	2.19 (0.25)*	2.51 (0.32)*
	AP	1.20 (0.25)	1.20 (0.24)	1.06 (0.23)$^{\ddagger,\S,\|}$	1.03 (0.20)$^{\ddagger,\S,\|}$	1.26 (0.29)†	1.41 (0.25)†	1.29 (0.33)†	1.34 (0.27)†	1.11 (0.32)†	1.45 (0.20)†
	ML	1.48 (0.45)	1.77 (0.60)	1.35 (0.42)	1.51 (0.42)	1.42 (0.42)	1.73 (0.54)	1.43 (0.31)	1.64 (0.44)	1.24 (0.36)	1.65 (0.47)
Lt Thigh	VT	2.39 (0.20)$^{\dagger,\|}$	2.31 (0.22)$^{\dagger,\|}$	2.24 (0.26)*	2.17 (0.24)*	2.19 (0.36)	2.26 (0.31)	2.17 (0.29)	2.13 (0.31)	2.16 (0.21)*	2.19 (0.36)*
	AP	1.32 (0.20)$^{\ddagger,\S,\|}$	1.22 (0.21)$^{\ddagger,\S,\|}$	1.20 (0.22)$^{\ddagger,\S,\|}$	1.16 (0.20)$^{\ddagger,\S,\|}$	1.43 (0.31)*,†	1.46 (0.15)*,†	1.43 (0.31)*,†	1.29 (0.38)*,†	1.49 (0.18)*,†	1.47 (0.26)*,†
	ML	1.42 (0.21)	1.46 (0.21)	1.31 (0.27)	1.36 (0.21)	1.34 (0.32)	1.49 (0.15)	1.42 (0.20)	1.36 (0.25)	1.50 (0.29)	1.54 (0.16)
Lt Shank	VT	2.13 (0.23)†	2.27 (0.27)†	2.08 (0.15)*,‡	2.06 (0.26)*,‡	2.17 (0.27)†	2.40 (0.29)†	2.07 (0.30)	2.29 (0.31)	2.08 (0.20)	2.10 (0.34)
	AP	1.28 (0.23)$^{\ddagger,\S,\|}$	1.30 (0.20)$^{\ddagger,\S,\|}$	1.15 (0.18)$^{\ddagger,\S,\|}$	1.21 (0.25)$^{\ddagger,\S,\|}$	1.39 (0.29)*,†	1.66 (0.22)*,†	1.46 (0.30)*,†	1.64 (0.26)*,†	1.61 (0.20)*,†	1.56 (0.32)*,†
	ML	1.46 (0.16)	1.27 (0.25)	1.36 (0.12)	1.15 (0.22)	1.39 (0.16)	1.38 (0.26)	1.32 (0.20)	1.34 (0.26)	1.35 (0.14)	1.30 (0.27)

Values: Mean (Standard Deviation)

*: Significant difference from very slow condition at $p < 0.05$, †: Significant difference from slow condition at $p < 0.05$, ‡: Significant difference from preferred condition at $p < 0.05$, §: Significant difference from fast condition at $p < 0.05$, ‖: Significant difference from very fast condition at $p < 0.05$.

Table 2-3 Z-values of the elderly in vertical (VT), antero-posterior (AP), and medio-lateral (ML) directions for each speed conditions.

		very low		low		preferred		fast		very fast	
		Men	Women	Men	Women	Men	Women	Men	Women	Men	Women
Head**	VT¶	2.79 (0.21)‖	2.99 (0.26)‖	2.68 (0.20)	2.85 (0.23)	2.73 (0.29)	2.93 (0.26)	2.58 (0.26)	3.00 (0.21)	2.46 (0.25)*	2.84 (0.27)*
	AP¶	1.96 (0.51)‖	1.22 (0.66)‖	1.71 (0.55)	1.32 (0.80)	1.60 (0.74)	1.18 (0.70)	1.52 (0.62)	1.49 (0.74)	1.39 (0.54)*	1.14 (0.54)*
	ML¶	1.44 (0.43)‖	1.46 (0.27)‖	1.41 (0.38)§	1.29 (0.28)§	1.54 (0.45)	1.60 (0.47)	1.50 (0.24)†	1.71 (0.23)†	1.55 (0.38)*	1.56 (0.42)*
Lumbar**	VT¶	2.71 (0.34)†,§,‖	3.01 (0.43)†,§,‖	2.61 (0.23)*	2.88 (0.23)*	2.69 (0.45)	3.06 (0.25)	2.47 (0.43)*	2.86 (0.39)*	2.42 (0.28)*	2.89 (0.27)
	AP¶	1.42 (0.31)	1.77 (0.58)	1.32 (0.25)	1.73 (0.38)	1.49 (0.30)	1.99 (0.41)†	1.37 (0.26)	1.85 (0.46)	1.35 (0.27)	1.88 (0.32)
	ML¶	1.09 (0.39)	1.21 (0.25)	0.92 (0.25)‡,‖	1.15 (0.22)‡,‖	1.12 (0.39)†	1.38 (0.34)†	1.07 (0.37)	1.29 (0.32)	1.06 (0.18)†	1.39 (0.32)†
Rt Hip**	VT¶	2.84 (0.38)†,‖	3.14 (0.13)†,‖	2.78 (0.18)*,‡	2.93 (0.18)*,‡	2.82 (0.33)†	3.02 (0.30)†	2.74 (0.35)	2.96 (0.17)	2.52 (0.26)*	2.83 (0.20)*
	AP¶	1.29 (0.30)†,‖	1.60 (0.29)†,‖	1.13 (0.29)*,‡	1.40 (0.24)*,‡	1.31 (0.37)†	1.67 (0.23)†	1.27 (0.35)	1.68 (0.24)	1.18 (0.30)*	1.66 (0.19)*
	ML	1.86 (0.49)†,‖	2.15 (0.60)†,‖	1.69 (0.47)*,‡	1.86 (0.60)*,‡	1.81 (0.48)†	2.01 (0.50)†	1.63 (0.51)	2.04 (0.55)	1.53 (0.54)*	1.94 (0.46)*
Rt Thigh	VT¶	2.66 (0.25)†,§,‖	2.86 (0.34)†,§,‖	2.41 (0.18)*	2.63 (0.27)*	2.70 (0.23)	2.71 (0.22)	2.46 (0.23)*	2.69 (0.27)*	2.26 (0.44)*	2.63 (0.18)*
	AP	1.25 (0.16)†,§	1.35 (0.18)†,§	1.08 (0.20)*,‡,§,‖	1.19 (0.14)*,‡,§,‖	1.52 (0.13)*	1.38 (0.18)*	1.41 (0.26)*,†	1.50 (0.16)*,†	1.42 (0.32)†	1.51 (0.18)†
	ML¶	1.14 (0.14)	1.36 (0.23)	0.95 (0.16)‡,§,‖	1.26 (0.23)‡,§,‖	1.28 (0.17)†	1.45 (0.20)†	1.13 (0.19)†	1.45 (0.19)†	1.14 (0.24)†	1.41 (0.17)†
Rt Shank	VT¶	2.34 (0.19)†	2.41 (0.34)†	2.19 (0.17)*,‡,§	2.27 (0.21)*,‡,§	2.42 (0.18)†,‖	2.50 (0.20)†,‖	2.35 (0.16)	2.49 (0.21)	2.25 (0.22)*,‡	2.43 (0.19)*,‡
	AP¶	1.44 (0.21)‡,§,‖	1.33 (0.18)‡,§,‖	1.33 (0.22)‡,§,‖	1.29 (0.21)‡,§,‖	1.69 (0.30)*,†	1.66 (0.14)*,†	1.75 (0.21)*,†	1.79 (0.22)*,†	1.76 (0.30)*,†	1.81 (0.22)*,†
	ML¶	1.31 (0.26)	1.12 (0.16)	1.17 (0.22)‡,§,‖	1.04 (0.14)‡,§,‖	1.41 (0.29)†	1.27 (0.18)†	1.33 (0.26)†	1.35 (0.21)†	1.39 (0.23)†	1.34 (0.15)†

(continued on next page)

Values: Mean (Standard Deviation)

*: Significant difference from very slow condition at $p < 0.05$, †: Significant difference from slow condition at $p < 0.05$, ‡: Significant difference from preferred condition at $p < 0.05$, §: Significant difference from fast condition at $p < 0.05$, ‖: Significant difference from very fast condition at $p < 0.05$, ¶: Significant difference between the elderly and the young, **: Significant difference between the elderly women and the young women.

Table 2-3 (CONT'D) Z-values of the elderly in vertical (VT), antero-posterior (AP), and medio-lateral (ML) directions for each speed conditions.

		very low		low		preferred		fast		very fast	
		Men	Women	Men	Women	Men	Women	Men	Women	Men	Women
Lt Hip	VT¶	2.81 (0.28)†,‖	3.00 (0.27)†,‖	2.60 (0.21)*	2.81 (0.19)*	2.67 (0.32)	2.96 (0.19)	2.52 (0.23)	2.87 (0.17)	2.41 (0.21)*	2.79 (0.09)*
	AP¶	1.18 (0.26)	1.43 (0.20)	1.05 (0.23)‡,§,‖	1.28 (0.28)‡,§,‖	1.32 (0.27)†	1.61 (0.18)†	1.30 (0.30)†	1.58 (0.24)†	1.25 (0.27)†	1.67 (0.16)†
	ML	1.65 (0.43)	1.67 (0.37)	1.45 (0.41)	1.50 (0.31)	1.63 (0.37)	1.69 (0.35)	1.46 (0.34)	1.65 (0.24)	1.42 (0.29)	1.62 (0.25)
Lt Thigh**	VT¶	2.61 (0.31)†,‖	2.71 (0.44)†,‖	2.45 (0.15)*	2.58 (0.33)*	2.59 (0.23)	2.62 (0.25)	2.45 (0.21)	2.61 (0.32)	2.34 (0.17)*	2.46 (0.41)*
	AP	1.12 (0.17)‡,§,‖	1.26 (0.23)‡,§,‖	1.02 (0.11)‡,§,‖	1.20 (0.13)‡,§,‖	1.28 (0.13)*,†	1.47 (0.17)*,†	1.38 (0.17)*,†	1.52 (0.21)*,†	1.27 (0.31)*,†	1.45 (0.35)*,†
	ML	1.38 (0.17)	1.43 (0.27)	1.20 (0.10)	1.34 (0.21)	1.42 (0.20)	1.50 (0.24)	1.35 (0.19)	1.52 (0.26)	1.23 (0.29)	1.43 (0.32)
Lt Shank	VT¶	2.40 (0.27)†	2.57 (0.28)†	2.20 (0.13)*,‡	2.32 (0.14)*,‡	2.44 (0.26)†	2.63 (0.18)†	2.24 (0.25)	2.59 (0.20)	2.24 (0.22)	2.52 (0.19)
	AP¶	1.47 (0.30)‡,§,‖	1.38 (0.23)‡,§,‖	1.32 (0.19)‡,§,‖	1.28 (0.21)‡,§,‖	1.66 (0.26)*,†	1.71 (0.23)*,†	1.67 (0.31)*,†	1.83 (0.15)*,†	1.74 (0.29)*,†	1.86 (0.17)*,†
	ML¶	1.60 (0.27)	1.41 (0.24)	1.44 (0.26)*,‡	1.29 (0.25)	1.62 (0.33)	1.50 (0.21)	1.47 (0.28)	1.50 (0.22)	1.52 (0.29)	1.48 (0.24)

Values: Mean (Standard Deviation)

*: Significant difference from very slow condition at $p < 0.05$, †: Significant difference from slow condition at $p < 0.05$, ‡: Significant difference from preferred condition at $p < 0.05$, §: Significant difference from fast condition at $p < 0.05$, ‖: Significant difference from very fast condition at $p < 0.05$, ¶: Significant difference between the elderly and the young, **: Significant difference between the elderly women and the young women.

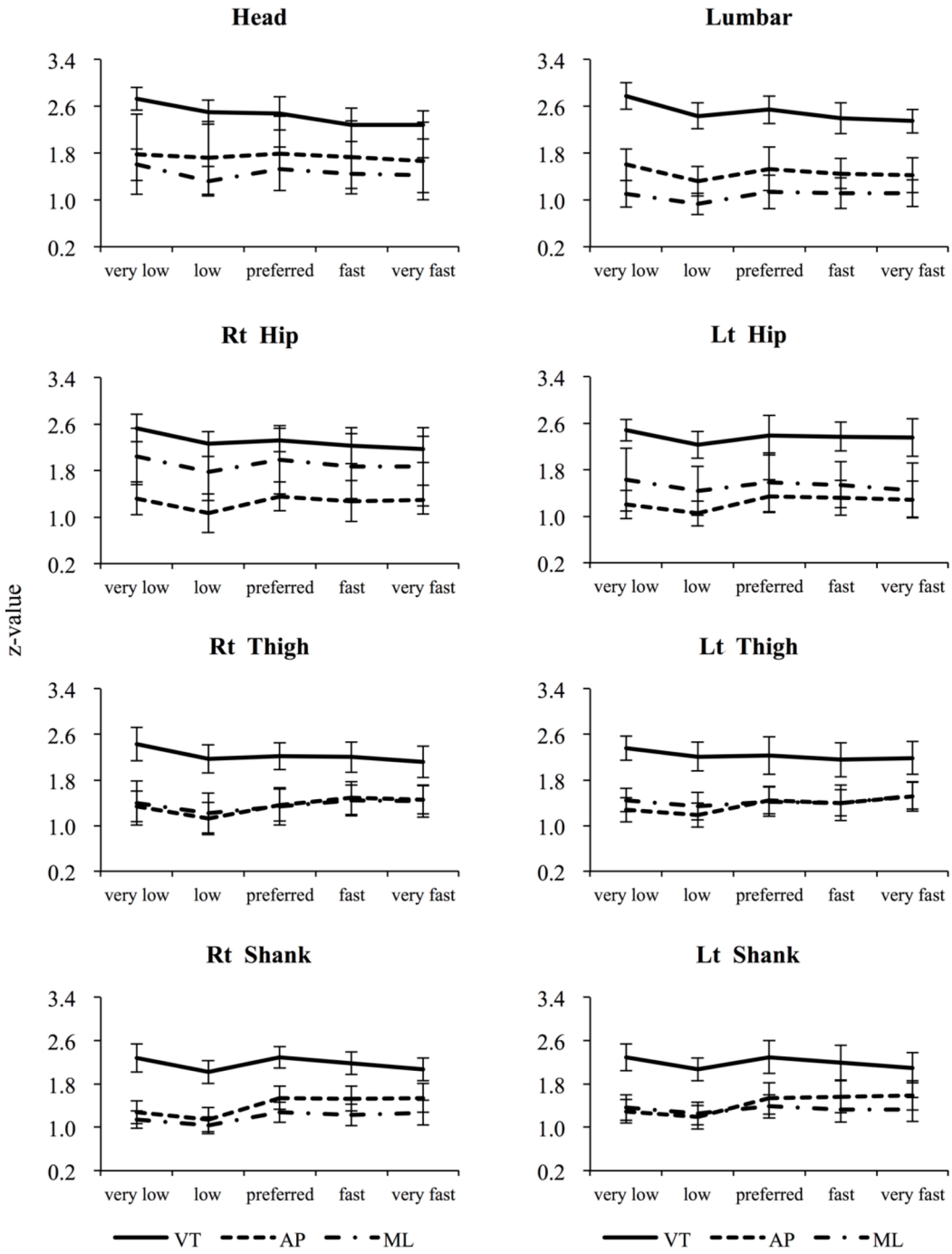


Figure 2-4 Effect of walking speeds on the z-value in vertical (VT), antero-posterior (AP), and medio-lateral (ML) direction of the young. Error bars span ± one standard deviation.

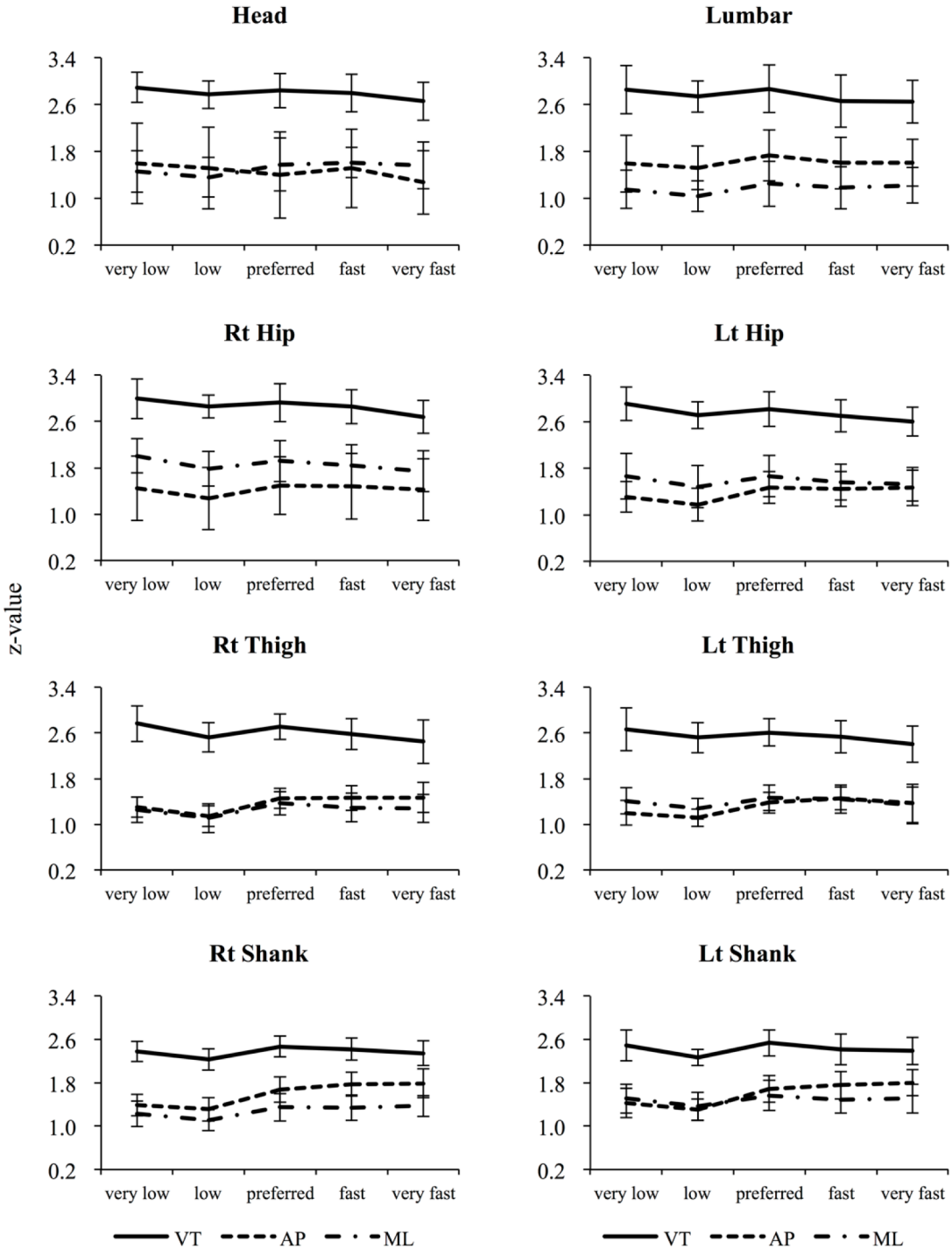


Figure 2-5 Effect of walking speeds on the z-value in vertical (VT), antero-posterior (AP), and medio-lateral (ML) direction of the elderly. Error bars span ± one standard deviation.

2.4. Discussions

This study examined the differences in the stride-to-stride acceleration variability by simultaneously evaluating acceleration of the head, lumbar, and lower extremities as a function of walking speed, and clarified whether the effect of walking speed on this acceleration variability differs among the body parts in the elderly and the young. This study revealed that the effects of walking speed on the stride-to-stride acceleration variability of all body parts of both the elderly and the young were similar. In addition, the stride-to-stride acceleration variability of the elderly was trend to be smaller than the value for the young, especially in vertical direction. Therefore, the hypotheses that the variability of accelerations of lower extremity of the elderly is easy to be affected by the walking speed and the variability of accelerations of lower extremity of the elderly is larger than the young are not supported.

Variability of vertical acceleration pattern of the lumbar and bilateral hip and thigh was smaller at very slow walking speed than low and very fast speeds, and that of bilateral shank was smaller at very slow walking speed than low speed. Vertical acceleration reflects the degree of support during walking. This result indicated that support pattern was more stable in very low speed condition. Previous studies reported that the variability of spatiotemporal parameters become a U-shaped pattern with increasing walking speed (Jordan et al, 2003; Danion et al., 2003). The result of this study did not consist with previous studies. Extra muscle activity occurs with decreasing walking speed for postural stability (den Otter et al., 2004). Vertical acceleration pattern

of the lumbar and lower extremity may be too stable by large muscle activity during very slow walking.

The effect of walking speed on the vertical acceleration variability on the left lower extremity was smaller than right lower extremity. Lower extremities have functional asymmetry, and non-dominant limb contributes more to support (Sadeghi et al., 2000). The subjects of this study were all right-handed. Since the left lower extremity of the study subject had the role of more support, the vertical acceleration variability on the left lower extremity might have been less influence on walking speed compared to right lower extremity. Moreover, the vertical acceleration variability at the all body parts was clearly smaller than AP or ML direction. Vertical control of acceleration was less affected by age and task condition (Asai, et al., 2013). These findings suggested that vertical acceleration of whole-body might be strictly controlled so as to stabilize the walking behavior.

On the bilateral thigh and shank, the influence of walking speed on the AP acceleration variability was observed. Z-values of these in very low and low condition were significantly smaller than the values for preferred, fast and very fast condition. AP accelerations reflect the degree of braking and propulsion and leg swing during walking. These are generated by the flexion-extension movement of the lower extremity. Plotnik et al. (2013) reported that bilateral coordination of left–right stepping was deteriorated when humans intentionally walk slowly. Because the walking speed effect for antero-posterior acceleration variability was the largest in the thigh and shank, the timing of step might have been controlled by the movement of the lower extremity.

Since antero-posterior acceleration variability of bilateral thigh and shank was larger in slower walking speed, therefore, the adjustment of stride by the lower extremity was not constant when walking slowly. On the other hand, the acceleration variability of upper body was small regardless of the change in walking speed. In AP direction, variability in the accelerations of the upper body and the lower extremity was exhibited characteristics unique to each body part.

Variability of ML acceleration pattern on the head, lumbar, and right lower extremity was clearly affected by the walking speed. Variability in this direction of head and lumbar tended to be larger at very slow walking speed, and variability of the right lower extremity tended to be smaller at slow walking speed. On the other hand, ML acceleration variability on the left lower extremity was not affected by the walking speed. Therefore, there is laterality in the effect of walking speed on ML acceleration variability. It has been reported that bilateral lower extremities have also functional asymmetry on lateral control as well as support, and ML ground reaction force during walking was mostly controlled by the left lower extremity (Matsuzaka et al., 1985). Since the left lower extremity had the role of control of lateral balance, ML acceleration variability on the left lower extremity might have been affected less by walking speed. On the other hand, ML acceleration variability on the right lower extremity was affected by walking speed, and became large when walking slowly. Because the elderly had instability in lateral direction related to the risk of fall (Rogers and Mille, 2003), it is important for the elderly to stably in lateral balance during walking. Therefore, it is

necessary for prevention of fall in the elderly to reduce the effect of walking speed on ML acceleration variability of the right lower extremity when walking slowly.

Walking speed was hardly to effect on acceleration variability of the head both the elderly and the young. Movement variability is thought to be a function of neurological integration and feedback processing of numerous sensory inputs that occur during the generation of each gait cycle (Chau et al., 2005). The head and eyes exhibit compensatory rotations in response to perturbations generated by the lower extremity movements to maintain stable head position and gaze (Hirasaki et al., 1999). Moreover, the vestibular system detects motion of the head, and stabilizes the gaze by the vestibuloocular reflex (Mitchell et al., 2013). The acceleration variability of the head was less affected by the walking speed and the gaze was stabilized, because acceleration from pelvis to head is absorbed by the spinal column (Kavanagh, 2004). As a consequence, this study revealed that there was body parts-specific effect of walking speed on stride-to-stride acceleration variability in each direction.

The elderly decreased the acceleration variability of all body parts in vertical direction, although there was not age-related change in the effect of walking speed on the acceleration variability. In AP and ML directions, the acceleration variability of some body parts was also smaller than the young. Mature motor skills and healthy states are associated with an optimal amount of movement variability that reflects the adaptability of the underlying control system, and a persistent lack of movement variability in the presence of changing task demands or environmental conditions may indicate rigid and inflexible motor behaviors with limited adaptability (Stergiou et al.,

2006). Previous study reported that the elderly significantly greater variability for trunk angle, step length and stride time (Kang and Dingwell, 2008), and greater variability of spatiotemporal parameters was associated with risk of fall (Menz et al., 2003b; Callisaya et al., 2010; Toebes M et al. 2012). By analysis using accelerometer, Moe-Nilssen (2005) reported that the frail elderly people had lower medio-lateral, but higher vertical and antero-posterior trunk variability than the fit ones. On the other hand, the acceleration variability of some body parts tended to be smaller than the young in this study. The results of this study were not consistent with previous studies, because the present study compared between the elderly and the young, and analyzed using the stride-to-stride acceleration variability of 10 strides of the head, lumbar, and lower extremity. This result suggested that the elderly performed more repeatable or stiff walking control that is likely to reduce sensory feedback and the ability to adjust for changing task demands or environmental conditions, because they had smaller acceleration variability of all body parts than the young regardless of changing walking speed.

Women were greater age-related alterations of the stride-to-stride acceleration variability than men. However, it was not possible to find a certain tendency in this study. Nevertheless, this is the first study to examine gender differences of the elderly in variability during walking. There were gender differences in body structure (Iglic et al., 2001; Cho et al., 2004) and muscle function (Lindle et al., 1997). Further studies measuring structural characteristics, body function and walking variability between

gender as well as age are needed to further the understanding between structure, function and control.

2.5 Summary

In this chapter, the objectives were to reveal the stride-to-stride acceleration variability through simultaneous evaluation of the acceleration of the head, lumbar, and lower extremity, and to clarify whether the effect of walking speed on the acceleration variability was different between the body parts in the elderly and the young. Moreover, the differences in the stride-to-stride acceleration variability of each body parts between the elderly and the young were evaluated. Twenty community-dwelling elderly people and 20 younger people participated in this study.

The results are as follows: 1) vertical acceleration might be strictly controlled so as to stabilize the walking behavior despite the effect of walking speed, 2) AP acceleration variability in very low and low condition smaller than the values for other conditions on the bilateral thigh and shank, 3) variability of ML acceleration pattern on the head, lumbar, and right lower extremity was clearly affected by the walking speed, and there is laterality in the effect of walking speed on ML acceleration variability, 4) the effects of walking speed on the stride-to-stride acceleration variability of all body parts of the elderly and the young were similar, 5) the stride-to-stride acceleration variability of the elderly was trend to be smaller than the value for the young.

In conclusion, the stride-to-stride acceleration variability of the lower extremity was affected when walking slowly both the elderly and the young. Additionally, the elderly exhibited more repeatable and rigid acceleration patterns regardless of changing walking speed.

Chapter 3

Control of ground reaction force during walking by the hip, knee and ankle joint moments

This chapter describes the control of the ground reaction force by the lower extremity in the elderly and the young. The aims are to find out the differences in the relationships between the vertical and antero-posterior ground reaction forces, and the lower extremity joint moment during walking between the elderly and the young. Forty elderly people, 65 years old or older, and 40 young people, aged 20 to 29 years participated in the experiment.

3.1 Purpose

The purpose of this study is to examine the differences of the relationships between the vertical and antero-posterior ground reaction force and the hip, knee, and ankle joint moments during walking in the elderly and the young individuals. It was hypothesized that the relation between the joint moments and these ground reaction forces differed

between the elderly and the young, and there was an age-related change in the strategy of walking.

3.2 Methods

3.2.1 Subjects

Forty community-dwelling elderly people, 65 years old or older, and 40 young people, aged 20 to 29 years, participated in this study. There were 20 men and 20 women in each group. The exclusion criteria included neurologic disorders, osteoarthritis, rheumatic arthritis, joint pain affecting walking, and a history of surgery in the lower extremities or spine. All procedures were approved by the Hiroshima International University Human Research Ethics Committee and all participants gave their written, informed consent prior to enrollment.

3.2.2 Experimental set up

Experimental set up of this study is shown in **Figure 3-1**. In the walking trials, all the participants walked barefoot to the end of a 7m walkway at a self-selected preferred walking speed. Data from 3 walking trials were collected. A 3D motion analysis system with 8 infrared cameras (VICON MX; Vicon Motion Systems; Oxford, UK) and 8 force plates (AMTI; Watertown, MA, USA) was used to record kinematic and kinetic data at sampling frequencies of 100 and 1,000 Hz, respectively. The force plate layout was designed to measure the ground reaction force of each limb using four force plates

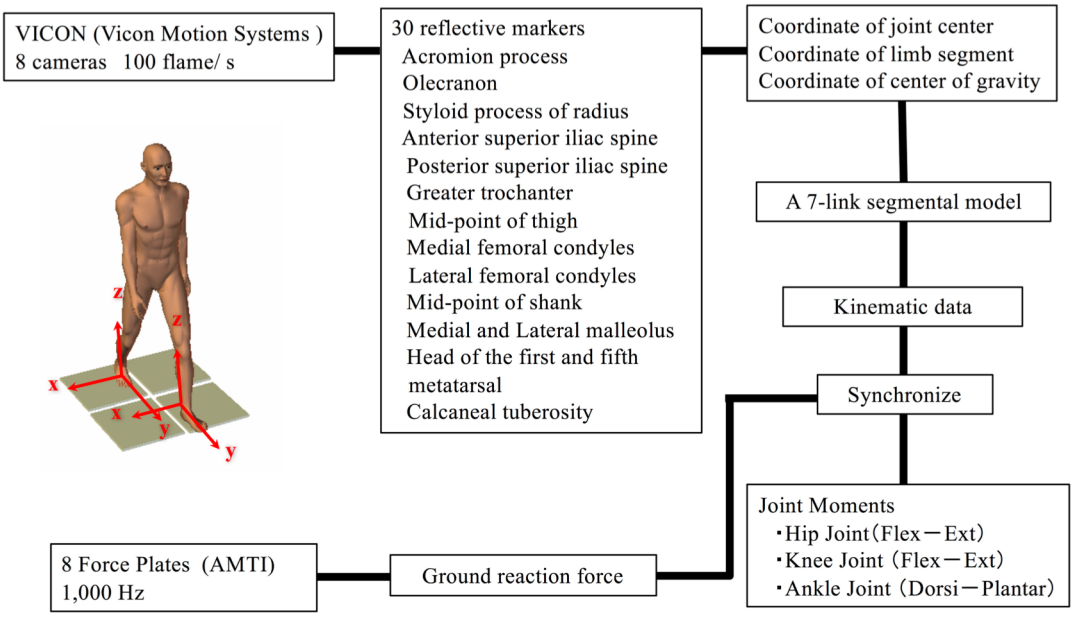


Figure 3-1 Experimental set up of this study.

arranged in the longitudinal direction. Participants were instructed to walk such that in each step they placed a foot on the left or right force plate, avoiding stepping on two force plates simultaneously.

A total of 30 reflective markers were placed bilaterally, over the following landmarks of each participant: acromion process, olecranon, styloid process of the radius, anterior superior iliac spine, posterior superior iliac spine, greater trochanter, medial and lateral femoral condyles, mid-point between the greater trochanter and the lateral femoral condyles, medial and lateral malleoli, mid-point between the lateral knee joint line and the lateral malleolus, head of the first and fifth metatarsal, and the calcaneal tuberosity. These anatomical markers were used to construct anatomical coordinate systems for the pelvis, thigh, shank, and foot segments.

3.2.3 Data analysis

The flow of data analysis is shown in **Figure 3-2**. Kinematic and kinetic calculations were performed using commercial software, "BodyBuilder" (Vicon Motion Systems). The coordinates of each joint center were calculated following the description of a preceding study (Andriacchi et al., 1982). The joint center of the knee was determined at a point interpolated at a distance of 2.6% of the body height from each reflective marker on the lateral femoral condyles along the vector linking the medial and the lateral femoral condyles markers. The ankle joint center was determined at a point interpolated at a distance of 2% of the body height from each reflective marker on the lateral malleolus along the vector linking the medial and the lateral malleolus markers. The joint center of the metatarsophalangeal joint was similarly determined at a point interpolated at a distance of 2.3% of the body height from each reflective marker on the head of the fifth metatarsal joint along the vector linking the head of the first and fifth

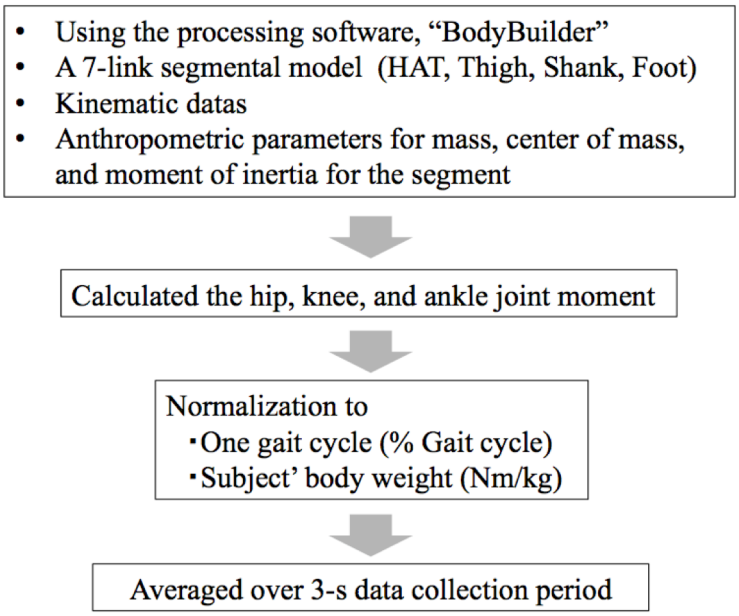


Figure 3-2 Flow of data analysis.

metatarsal markers. The joint center of the hip was determined as follows: first, I calculated a vector, which links the right and left greater trochanter markers. Second, the joint center of the hip was determined at a point interpolated at a distance of 18% of the vector norm from each reflective marker of the superior aspect of the greater trochanter along the vector.

Calculation methods of joint moment are provided at **Appendix A**. A 7-link segmental model was developed to calculate the kinematic and kinetic variables of the hip, knee, and ankle joints using the inverse dynamics technique of Davis et al. (1991) and Vaughan et al. (1992) Anthropometric parameters for mass, position of the center of mass, and moment of inertia for the segments were obtained from the report by Okada et al (1996). The hip, knee, and ankle joint moments using a local coordinate system with the origin at the joint center were calculated. In this study, each joint moment was normalized to the subject's body mass (Nm/kg).

Walking speed (m/s) was calculated using the center of gravity (COG) of the whole body, and averaged over a 3-s data collection period. Stride length (m) was measured as the antero-posterior distance between the left calcaneal tuberosity marker at one heel contact and at the next heel contact. Stride length was normalized to body height (%BH).

Initial contact was assumed to occur when the vertical reaction force exceeded 10 N, and toe off was assumed to occur in the first frame following the initial contact when the vertical reaction force fell below 10 N. To account for different builds, ground reaction force was normalized to each subject’s body mass (N/kg). First and second

peaks of the vertical component, and the minimum value between the two peaks, and the peak values of the antero-posterior component were extracted from the ground reaction force profiles (**Figure 3-3**). Peak values of each joint moment during the stance phase were extracted from the joint moment profiles (**Figure 3-3**).

3.2.4 Statistical analysis

A two-way analysis of variance was performed to examine the differences between the elderly and the young and the interactions between age and gender, with age, body height, body weight, body mass index (BMI), and spatiotemporal parameters. Two-way analysis of covariance was also performed to examine the difference between the elderly and the young, and the interactions between age and gender with the vertical and antero-posterior ground reaction force, and peak values of each joint moment using walking speed as the covariate.

A step-wise linear regression was performed to determine the joint moments that predicted the amplitude of the vertical and antero-posterior ground reaction force in the elderly men, elderly women, young men, and young women. Relationships among the peak values of the hip, knee, and ankle joint moments during the early stance phase and the first peak force, and the minimum value of the vertical component and the posterior component of ground reaction force were investigated. The relationships among the peak values of the hip, knee, and ankle joint moments during the late stance phase and the second peak force of the vertical component and the anterior component of ground reaction force were also investigated. The stepping-method criteria were: a

F-value of ≥0.05 for inclusion in the segmental model, and ≤0.10 for removal from the model.

Statistical significance was accepted for values of $p < 0.05$. All data were analyzed using SPSS 17.0 statistical software (SPSS Japan Inc., Tokyo, Japan).

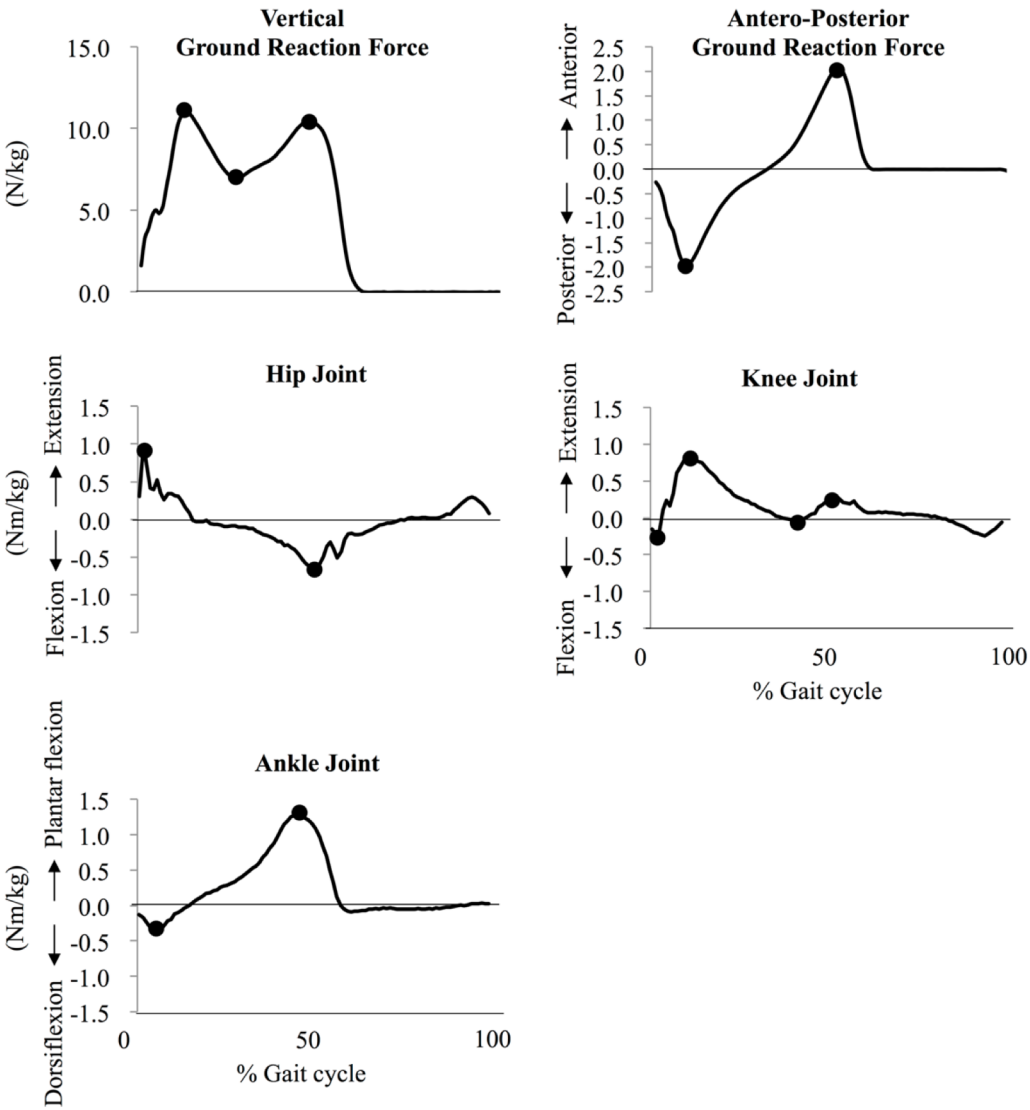


Figure 3-3 Graphical representations of the vertical and antero-posterior ground reaction forces and the joint moment of the hip, knee, and ankle during a gait cycle.

3.3 Results

The baseline demographic characteristics and spatiotemporal parameters of the study subjects are presented in **Table 3-1**. The elderly participants had significantly shorter body height and higher BMI than the young study participants, but there was no significant difference in their body masses.

Comparisons of the ground reaction force and peak values of the joint moment are presented in **Figure 3-4**. The elderly men and women had significantly slower walking speeds, with shorter stride lengths than the young participants, but there was no significant different in cadence between the two age groups. The second peak of the vertical ground reaction force exhibited a significant interaction between age and gender, and the elderly women had a significantly lower the second peak of the vertical ground reaction force than the young women. In addition, both anterior and posterior components of the ground reaction force were lower in the elderly participants. The anterior component of the ground reaction force had a significant interaction between

Table 3-1 Characteristics of the subject.

Variable	Elderly		Young	
	Men (n = 20)	Women (n = 20)	Men (n = 20)	Women (n = 20)
Age (years)	69.45 (4.80)*	70.70 (4.62)*	22.80 (2.80)	23.50 (2.70)
Height (m)	1.64 (0.07)*	1.49 (0.04)*	1.71 (0.07)	1.57 (0.05)
Weight (kg)	62.28 (9.78)	52.09 (6.20)	64.04 (8.34)	52.91 (9.92)
BMI †(kg/m^2)	23.06 (2.96)*	23.25 (2.69)*	21.74 (2.63)	21.21 (3.30)
Walking speed (m/s)	1.20 (0.16)*	1.15 (0.12)*	1.26 (0.12)	1.29 (0.10)
Stride length (%BH‡)	73.63 (6.79)*	73.59 (5.48)*	77.47 (5.64)	81.68 (5.69)

Value: mean (Standard deviation: SD)

*: Significant difference from the young, $p < 0.05$

†: BMI: Body Mass Index

‡: % Body Height

age and gender.

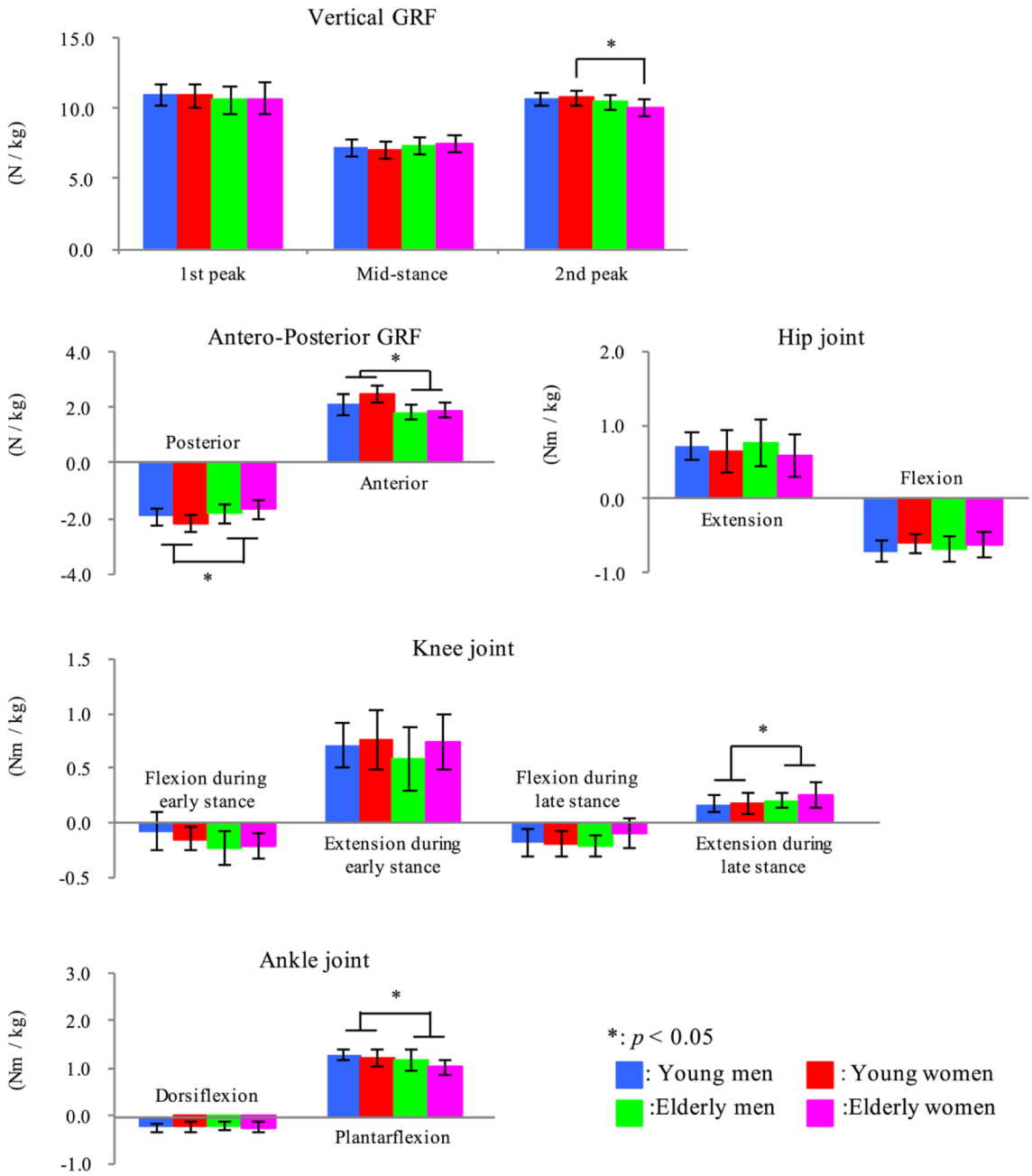


Figure 3-4 Comparison of vertical and antero-posterior ground reaction forces (GRF) and lower extremity joint moments. Error bars span ± one standard deviation.

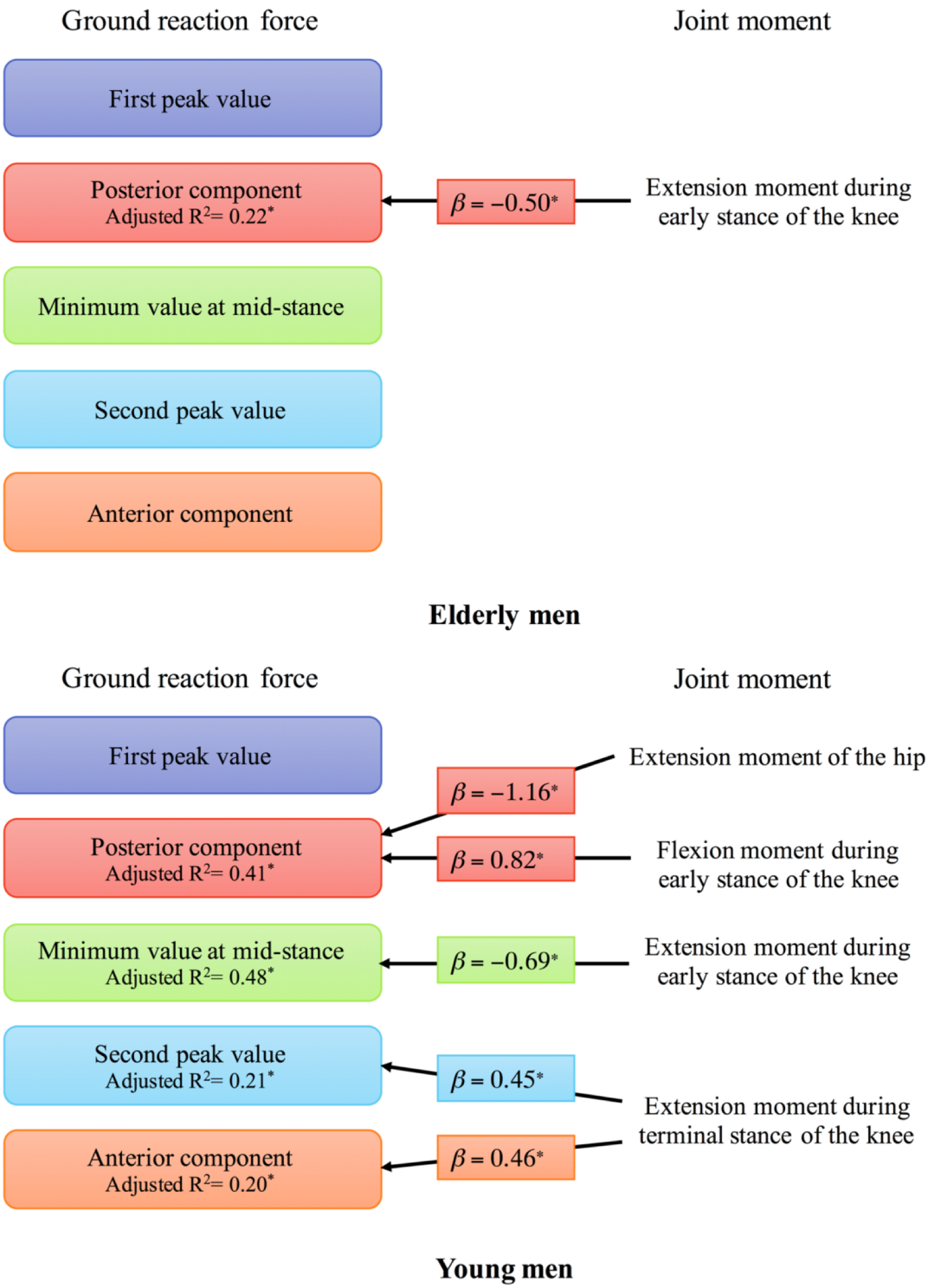


Figure 3-5 Models of the vertical and antero-posterior ground reaction forces by forward step-wise regression analysis of the elderly men and the young men. * indicates $p < 0.05$.

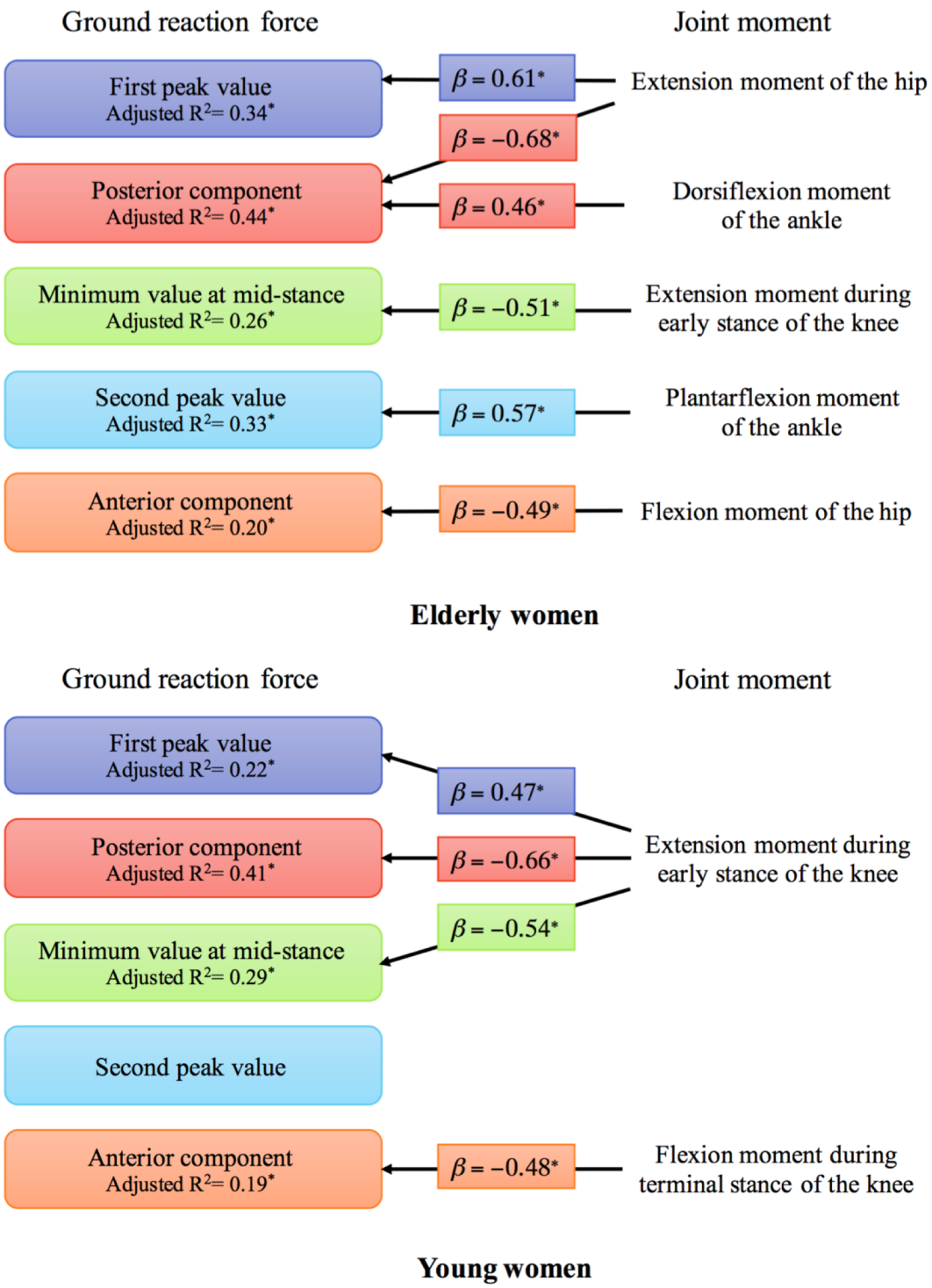


Figure 3-6 Models of the vertical and antero-posterior ground reaction forces by forward step-wise regression analysis of the elderly women and the young women. * indicates $p < 0.05$.

The plantar flexion moment of the ankle was lower in the elderly participants. On the other hand, the extension moment during the terminal stance phase of the knee was higher in the elderly participants. Although the flexion moment during the terminal stance phase of the knee exhibited a significant interaction between age and gender, there was no significant difference between the age groups, neither in men nor in women.

The results of step-wise linear regression analyses of the vertical and antero-posterior ground reaction forces are presented in **Figure 3-5** and **3-6**, and the joint moments related to the vertical and antero-posterior ground reaction forces are illustrated in **Figure 3-7**. In elderly men, extension moment during early stance of the knee was able to explain the results of the posterior component of the ground reaction force. Unfortunately, there was no statistically significant factors related to the other variables of the vertical and anterior component of the ground reaction force.

For young men, both extension moment of the hip, and flexion moment during early stance of the knee were related to the posterior component of the ground reaction force, and the knee extension moment during the late stance phase was related to the second peak of the vertical component and the anterior component of the ground reaction force.

On the other hand, for young women, the knee extension moment during the early stance phase was related to the first peak of the vertical component and the posterior component of the ground reaction force. Moreover, flexion moment during

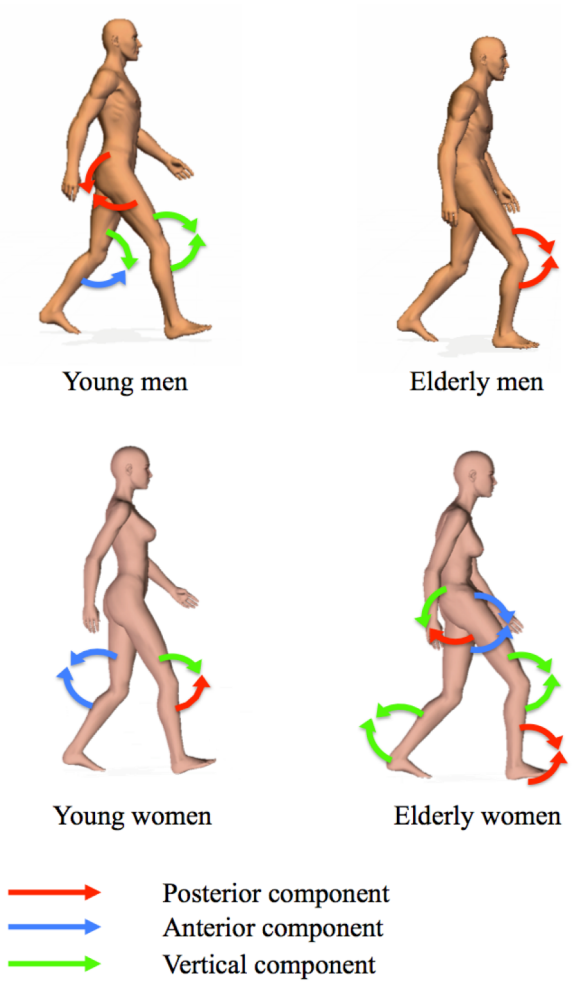


Figure 3-7 The joint moments that can predict the amplitude of the vertical and antero-posterior ground reaction force in each group.

terminal stance of the knee was related to the anterior component of the ground reaction force.

In contrast, for the group of elderly women, the extension moment of the hip was related to the first peak of the vertical and posterior component of the ground reaction forces. In addition, the dorsiflexion moment of the ankle was also related to the posterior component of the ground reaction force. The plantarflexor moment was

related to the second peak of vertical ground reaction force. With regard to the anterior component of the ground reaction force, the flexion moment of the hip exhibited explanatory power in the model.

For common in the young men and women, and elderly women groups, the knee extension moment during the early stance phase had a negative relationship with the minimum value of the vertical ground reaction force at mid-stance.

3.4 Discussion

This study examined the relationships between the vertical and antero-posterior component of the ground reaction force and hip, knee, and ankle joint moments during walking in the elderly and the young. The first peak and the minimum value at mid-stance of the vertical ground reaction force were not significantly different between the elderly and the young. On the other hand, the second peak of the vertical ground reaction force of the elderly women was significantly lower than that of the young women. Yamada and Maie (1988) reported that elderly people have lower first peak and second peak forces, and a higher minimum value at mid-stance than young people. On the other hand, Larish et al. (1988) reported that there was no significant difference in the first peak force, but a significant difference in the second peak of the vertical ground reaction force. Because the amplitude of the peak vertical ground reaction force is affected by cadence rather than stride length (Martin et al., 1992), there may be no difference between the elderly and young people in the first peak of the

vertical ground reaction force. Both the first and second peaks of the vertical ground reaction force correspond to the upward acceleration of the body center of mass (Jacobs et al., 1972). The elderly women in this study had low upward acceleration of body center of mass at push-off. This may increase the load on the front limb when producing upward acceleration during the double support phase of elderly women.

The antero-posterior ground reaction forces in the elderly both men and women were significantly decreased than the young. Results in this study were consistent with previous study (Yamada and Maie, 1988). The antero-posterior ground reaction forces act as a braking force and propulsive force to body center of mass, respectively, and relate stride length and walking speed (Yamada and Maie, 1988). In the elderly, because the antero-posterior ground reaction forces were decreased, stride length was shortened, and walking speed was reduced.

The plantar flexion moment of the ankle in the elderly was smaller by approximately 14.5 percent than that of the young people. Therefore, the plantar flexor muscles of the elderly exhibited an age-related functional decline. Young people used 40% of their maximum plantar flexor moment during walking (Stefanyshyn et al., 1998). The plantar flexor during walking exerts the largest force. Therefore, it is important for the elderly to strengthen the ankle plantar flexor.

The knee extension moment during early stance had a negative relationship with the minimum value at mid-stance, and the knee extension moment during late stance had a positive relation with the second peak of the vertical ground reaction force of the young men. The knee extension moment during early stance had a positive

relation with the first peak force, and a negative relation with the minimum value at mid-stance of the ground reaction force of the young women. These results suggest that the vertical ground reaction force may be mainly controlled by knee extensor in young people. In addition, the knee joint moment during late stance was associated with the anterior component of the ground reaction force of the young men and women. The knee joint moment of young people is important for decreasing the reaction force from the ground to the body at mid-stance, and increasing the ground reaction force during weight acceptance and the push off phase. It is necessary to have muscle strength in order to increase the joint moment during walking. Brown et al. (1995) reported that there was a relationship between quadriceps strength and walking ability, especially in young people rather than in the elderly. Therefore, the knee joint function is most important for walking by young people.

On the other hand, the hip extension moment and the knee flexion moment during early stance were related with posterior component of the ground reaction force in young men, whereas only knee extension moment was related with this force in young women. These results suggest that the posterior component of the ground reaction force is generated by only knee moment in young women, whereas it is generated by a combination of hip and knee moment in young men. The posterior ground reaction force is responsible for shock absorption. In other words, young women may perform shock absorption depending on the knee joint. This thing may be associated with the onset of knee osteoarthritis in women.

In the elderly women of this study, the extension moment of the hip had a positive relationship with the first peak and the posterior component of the ground reaction forces. The elderly women had a large hip extension moment, and their ground reaction force during the early stance phase was relatively high. Although the knee joint moment was associated with the first peak of the vertical ground reaction force in the young women, the hip joint moment was associated with the first peak of the vertical ground reaction force in the elderly women. Weight acceptance for the body is provided by a combination of knee extensors and hip extensors during the early stance phase (Kepple et al., 1996). Elderly people have significantly weaker quadriceps and hamstrings muscle strength than young people (Moreland et al., 2004). While walking, moreover, elderly people had more forward trunk lean, and a large hip flexion angle during early stance (DeVita and Hortobagyi, 2000). Because the function of the muscles surrounding the knee joint reduces with aging in addition to age-related alternation in the attitude of walking, weight acceptance during the early stance phase of the elderly is accomplished by the hip joint. For weight acceptance in the initial stance, therefore, it is necessary for elderly women to develop a hip extensor.

The knee extension moment during the early stance phase had a negative relationship with the minimum value of the ground reaction force at mid-stance in the elderly women. This result was similar to the one found for the young women in this study. In order to reduce the mechanical load exerted on the body at mid-stance, it may be important for elderly women to strengthen the quadriceps muscles, and generate a knee extension moment during early stance.

The ankle plantar flexion moment had a positive relationship with the second peak of the ground reaction force of the elderly women. Winter (1980) reported that the ankle plantar flexion moment was the major contributor to support during push-off. The rate of decline in maximal voluntary isometric force with aging is greatest in the plantar flexors (Christ et al., 1992). It is probable that because the elderly women had a lower plantar flexion moment of the ankle than that of young women, the second peak of the vertical ground reaction force was smaller in the elderly women in this study. A preceding study reported the importance of plantar flexor muscles of the ankle during walking in the elderly (Graf et al., 2005). Therefore, it is important to strengthen the plantar flexor muscles of the ankle of elderly women in order to enhance support during late stance.

The knee joint moment was related to the anterior ground reaction force in young women. Kepple et al. (1997) reported that knee extension moment in addition to ankle plantar flexion moment contributed to forward acceleration. It follows that knee function may be important for propulsion during walking. On the other hand, the flexion moment of the hip had a positive relationship with the anterior ground reaction force of the elderly women. McGibbon (2003) reported the elderly might use hip flexion for the control of the body thereafter mid-stance and forward progression of leg into swing as a means of compensating for ankle plantar flexor. Results in this study were consistent with previous study (McGibbon, 2003). From these results, the elderly and young women in this study made use of different joints to control of the vertical and

antero-posterior ground reaction force. These results are mostly consistent with our hypothesis that there is an age-related change in the strategy of walking.

In elderly men, only extension moment during early stance of the knee had a positive effect on this force, whereas extension moment of the hip, and flexion moment during early stance of the knee was related in young men. These results suggest that the posterior component of ground reaction force is generated by only knee moment in the elderly, whereas it is generated by combination of hip and knee moment in the young men. It follows that the elderly and young men made use of different joints for production of the posterior component of the ground reaction force. The results may indicate that elderly men experience age-related decreases in the functioning of the hip. However, no factors related to the vertical component and the anterior component of ground reaction force of the elderly men were found. Because individual variation was greater among the elderly men, factors related to the vertical ground reaction force were found not to be significant, similar to the antero-posterior ground reaction force. The changes in physical function with aging are not equal in men and women (Lindle et al., 1997), and physical characteristics, lifestyle, and social background are different between men and women. Therefore, there are gender differences in the age-related changes of walking characteristics, and the elderly men make use of various joint moment patterns to the control of the vertical ground reaction force.

3.5 Summary

In this chapter, the objectives were to reveal the differences in the relationships between the vertical and antero-posterior ground reaction forces and the hip, knee, and ankle joint moments during walking between the elderly and the young. Forty community-dwelling elderly people and 40 young people, participated in this study.

The results are as follows: 1) In young men, the extension moment during early and late stance of the knee was related to the minimum value at mid-stance, and the second peak force and the anterior component of the ground reaction force, respectively. Extension moment of the hip was related to the posterior ground reaction force. 2) In elderly men, the extension moment during early of the knee was related to the posterior ground reaction force, although there were no factors related to the vertical ground reaction force. 3) In young women, the knee joint moment was related to the first peak force, and minimum value at mid-stance, and the antero-posterior component of the ground reaction forces. 4) In the elderly women, all of the hip, knee and ankle joint moments were related to the vertical and antero-posterior ground reaction forces.

In conclusion, the elderly and the young both men and women used the lower limb joints differently for the control of the vertical and antero-posterior ground reaction forces. In addition, there are gender differences in the age-related differences of the relationship between the ground reaction force and the lower extremity joint moment.

Chapter 4

Muscle control of lower extremity during walking

This chapter describes about the analysis of muscle control of lower extremity during walking using musculo-skeletal simulation in the elderly and the young. The aims are to examine the differences of the amplitude, timing and combination of muscle force during walking between the elderly and the young. Twenty elderly people, 65 years old or older, and 20 young people, aged 20 to 29 years participated in the experiment.

4.1 Purpose

The purpose of this study was to examine the age-related differences of the amplitude, timing and combination of muscle force during walking. Furthermore, the relationship between muscle modules and support and propulsive force during walking was also examined. It was hypothesized that there were the age-related alterations of muscle coordination during walking, and support and propulsive force corresponding to these muscle modules were reduced.

4.2 Methods

4.2.1 Subjects

Twenty community-dwelling elderly people, 65 years old or older and 20 younger people, aged 20 to 29 years participated in this study. There were 10 men and 10 women in each group. The exclusion criteria included neurologic disorders, osteoarthritis, rheumatic arthritis, joint pain affecting walking, and history of surgery in the lower extremities or spine. All participants gave their written, informed consent prior to enrollment.

4.2.2. Experimental set up

In the walking trials, all participants walked with barefoot to the end of a 7 m walkway at a self-selected preferred walking speed. A 3 D motion analysis system with 8 infrared cameras (VICON MX; Vicon Motion Systems; Oxford, UK) and 8 force plates (AMTI; Watertown, MA, USA) was used to record kinematic and kinetic data at sampling frequencies of 100 and 1,000 Hz, respectively. The force plate layout was designed so as to measure ground reaction force of each limb using four force plates arranged in the longitudinal direction. Participants were instructed to walk such that in each step they placed a foot on the left or right force plate, avoiding stepping on two force plates simultaneously. A total of 30 reflective markers were placed bilaterally, at the following landmarks on each participant: acromion process, olecranon, styloid process of radius, anterior superior iliac spine, posterior superior iliac spine, greater trochanter, medial and lateral femoral condyles, mid-point between greater trochanter and lateral femoral

condyles, medial and lateral malleolus, mid-point between lateral knee joint line and lateral malleolus, head of the first and fifth metatarsal, and calcaneal tuberosity.

4.2.3. Gait evaluation

Walking speed (m/s) was calculated from the position of center of gravity (COG) of the whole body. Stride length (m) was measured as the antero-posterior distance of the position of left calcaneal tuberosity marker at a heel contact and the next heel contact. Stride length was normalized to body height (%BH). Initial contact was assumed to occur when vertical reaction force exceeded 10 N, and toe off was assumed to occur in the first frame following the initial contact where the vertical reaction force was <10 N.

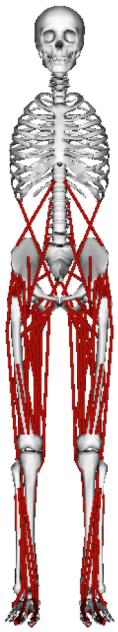


Figure 4-1 Musculoskeletal model.

4.2.4. Muscle force estimation

Muscle forces during walking were estimated using OpenSim 3.2 (Delp et al. 2007). The body was modeled as a 23 degree-of-freedom linkage actuated by 92 musculo-tendon actuators (**Figure 4-1**). The head, arms and torso were modeled as a single rigid body, which articulated with pelvis *via* a ball-and-socket back joint. Hip joint was modeled as ball-and-socket joints. Knee joint was modeled as a hinge joint. Ankle-subtalar complex was modeled as a universal joint. Metatarsal-phalangeal joint was modeled as a hinge joint. Muscle parameters and path geometries were based on the data reported by Delp et al. (1990). Optimal muscle fiber lengths and tendon slack lengths were scaled from a generic model to fit each subject using segment dimensions

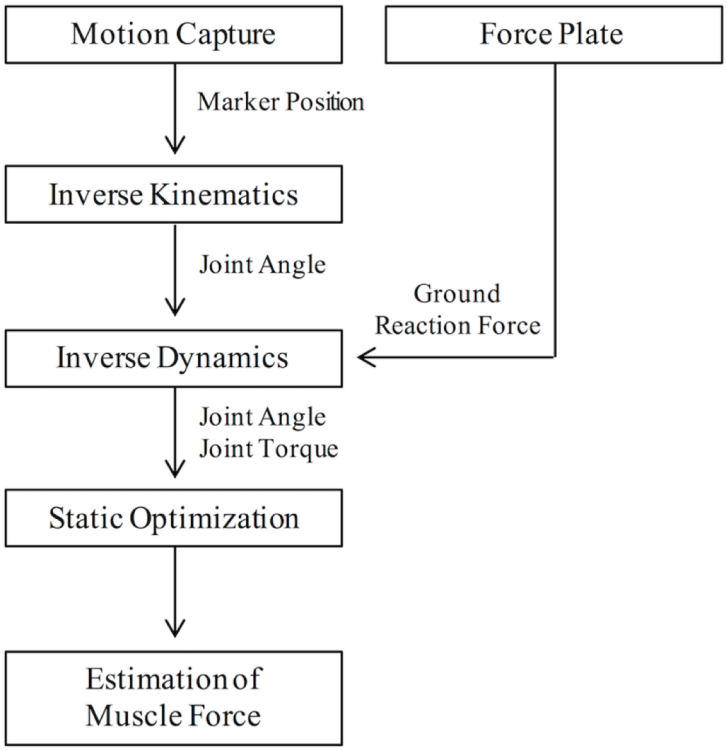


Figure 4-2 Flow of muscle force estimation.

as scaling factors.

Both ground reaction forces and marker trajectories were low-pass filtered with a fourth-order Butterworth filter (6 Hz). Data from the left leg were used in the analysis. OpenSim software (Delp et al. 2007) was used to generate and analyze simulations of walking using a single representative gait trial of each subject. The flow of muscle force estimation is shown in **Figure 4-2**. Inverse kinematics and inverse dynamics were subsequently applied, in conjunction with the measured marker trajectories and the ground reaction forces and moments, to obtain a set of dynamically consistent joint angles and moments. Muscle forces were calculated using static optimization toolbox into OpenSim. The static optimization algorithm is expressed as:

$$\sum_{m=1}^{n} [a_m f(F_m, l_m, v_m)] r_{m,j} = \tau_j \tag{3.1}$$

while minimizing the sum of squares of all muscle activations in the model (Crowninshield et al. 1978):

$$J = \sum_{m=1}^{n} (a_m)^2 \tag{3.2}$$

where n is the number of muscles in the model, a_m is the activation level of muscle m, F_m is its maximum isometric force, $f(F_m, l_m, v_m)$ is its force-length-velocity surface, $r_{m,j}$ is its moment arm about the jth joint axis, τ_j is the generalized force acting about the jth joint axis. Musculotendinous model for calculation of static optimization in OpenSim are provided at **Appendix B**. Gluteus maximus (GMAX), gluteus medius (GMED),

gluteus minimus (GMIN), iliopsoas (IL), rectus femoris (RF), vastii muscles (VAS), hamstrings (HAM), gastrocnemius (GAS), soleus (SOL) and tibialis anterior (TA) were analyzed. Each muscle force was normalized to the subject's body mass (N/kg) for the comparison between the elderly and the young people. Peak values during a gait cycle, and average values during stance and swing phase of each muscle force were extracted from the force profile.

4.2.5. Non-negative matrix factorization

Non-negative matrix factorization (NMF) algorithm (Lee and Seung, 1999; Ting and Macpherson, 2005) was used for the analysis of muscle synergy modules (**Figure 4-3**, **4-4**). To perform NMF, the data were arranged into an m×n data matrix for each subject, where m is the number of muscles and n is the time base (n = 101). NMF defines the

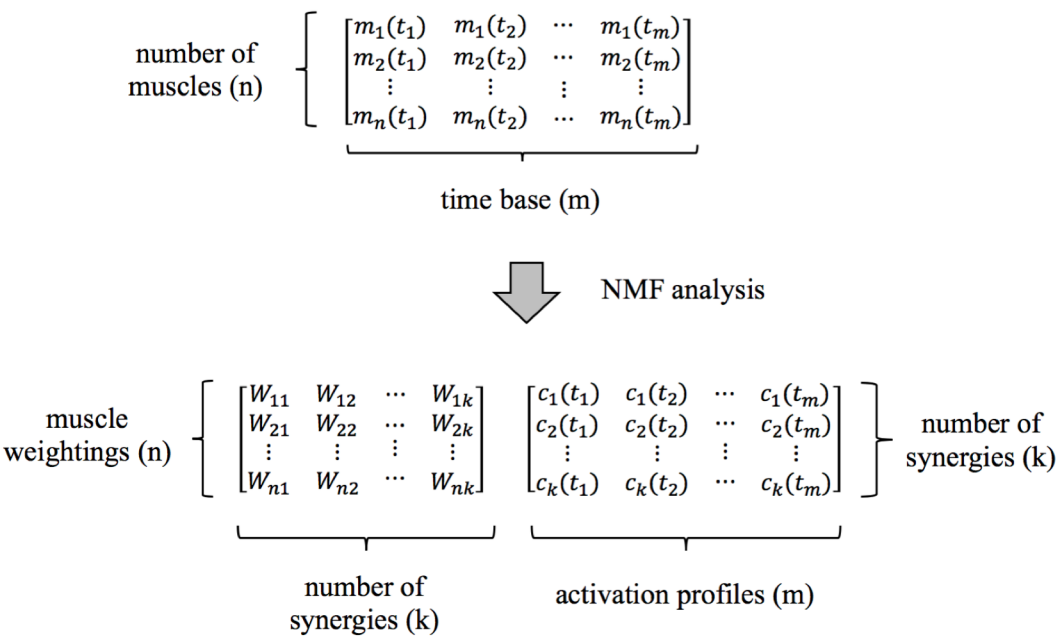


Figure 4-3 Organization of the data matrix using NMF analysis.

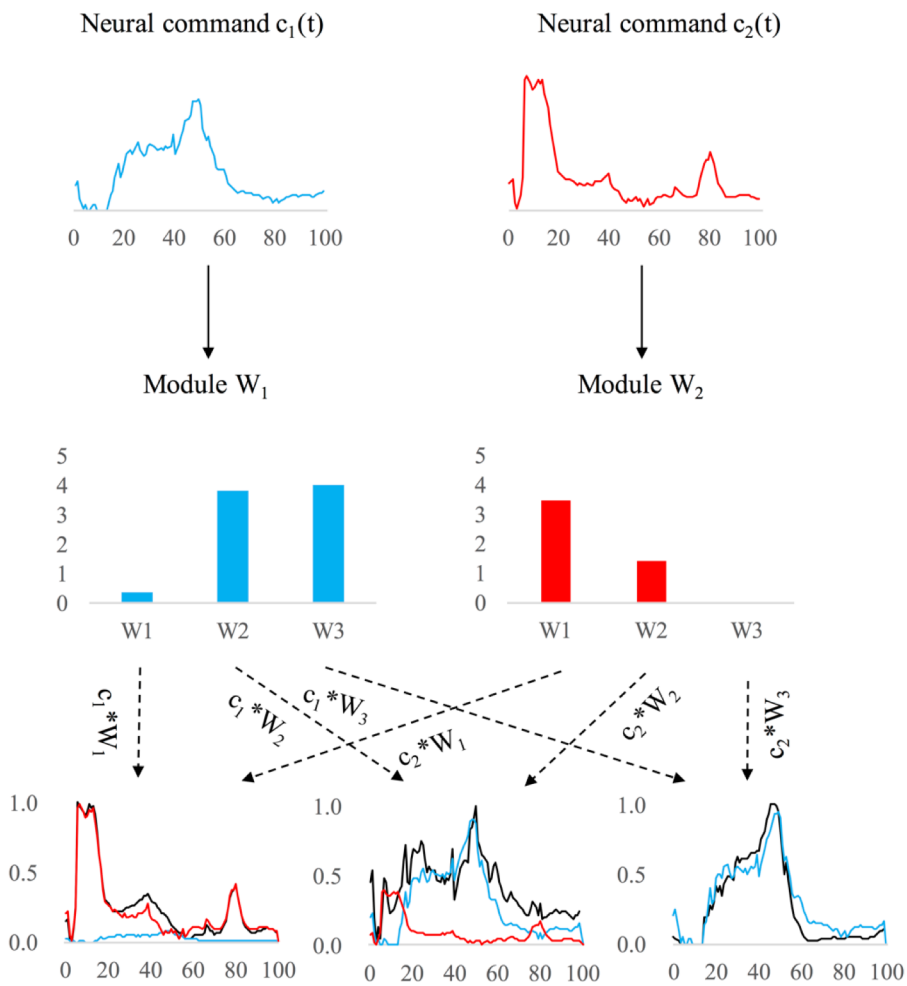


Figure 4-4 A schematic description of muscle control by modules. In this example, there are two modules and three muscles. These muscle activations can be represented in term of the lower-dimensional combination of muscle modules (W_i) and neural activation commands ($c_i(t)$).

muscle activation modules by populating two matrices: (1) a m×k matrix indicating the relative weighting of each muscle within each module and (2) a k×n matrix reflecting the activation timing profile of the module across the gait cycle, where k is the number of modules (Neptune et al., 2009) (see **Appendix C**). Muscle force values of each muscle were normalized to the maximum value for that muscle so that each value fits

between 0 and 1. To select the number of muscle synergies that could best reproduce the recorded data, 1-7 synergy matrices and synergy activation profile matrices from the muscle force data matrices obtained for each subject were extracted. Then, the goodness-of-fit between the original data (D_o) and the reconstructed data (D_r) matrices, to select the smallest number of muscle synergies that result in an adequate reconstruction of muscle responses were verified. The variability accounted for (VAF) as the ratio of the sum of squared error values to the sum of squared original dataset values [$VAF = 1 - (D_o - D_r)^2 / D_o^2$], which was based on the entire dataset was calculated. VAF was sensitive to both the shape and magnitude of the original and reconstructed data sets. Calculations of NMF were performed using built in function nnmf from Matlab R2013a (The MathWorks, Inc., Natick, MA, USA).

In order to evaluate the relationship between the age-related difference in muscle modules and support and propulsive forces during walking, the vertical and antero-posterior ground reaction forces corresponding to the peak value of the activation timing profiles of each module were identified. To account for different builds, ground reaction force was normalized to each subject's body mass (N/kg).

4.2.6. Statistical analysis

A three-way analysis of variance with the number of synergies, age, and gender as factors was performed in order to determine the number of contributing synergies. A two-way analysis of covariance was performed to examine the difference between the elderly and the young and the interaction between age and gender groups in muscle

forces and muscle weightings using walking speed as a covariate. The similarities of the activation timing profiles between the elderly and the young were analyzed by cross-correlation analysis in men and women, respectively. The level of statistical significance was set at $p < 0.05$. All data were analyzed with SPSS 17.0 statistical software (SPSS Japan Inc., Tokyo, Japan).

4.3. Results

The baseline demographic characteristics and spatiotemporal parameters of the study subjects are presented in **Table 4-1**, and the joint moments of the lower extremity are indicated in **Table 4-2**. The elderly participants were significantly shorter in body height and higher in BMI, but not significantly different in body mass, as compared with young participants. The elderly men and women had significantly slower walking speeds along with shorter stride lengths than the younger participants.

Table 4-1 Characteristic of subjects.

Variable	Elderly		Young	
	Men (n = 10)	Women (n = 10)	Men (n = 10)	Women (n = 10)
Age (years)	68.4 (3.0)*	69.1 (4.3)*	21.7 (2.1)	24.1 (2.9)
Height (m)	1.65 (0.07)*	1.50 (0.03)*	1.71 (0.07)	1.57 (0.04)
Weight (kg)	64.3 (10.0)	53.9 (7.8)	60.6 (5.1)	48.3 (6.5)
BMI †(kg/m^2)	23.5 (2.8)*	23.8 (3.4)*	20.6 (1.6)	19.4 (1.7)
Walking speed (m/s)	1.16 (0.12)*	1.15 (0.13)*	1.28 (0.14)	1.27 (0.07)
Stride length (%BH‡)	72.9 (5.7)*	73.1 (5.0)*	78.7 (5.0)	81.1 (3.7)

Value: mean (Standard deviation: SD)

* : Significant difference from the young at $p < 0.05$

† : BMI: Body Mass Index

‡ : % Body Height

Muscle force values are presented in **Figure 4-6**. Peak values of GMAX and GMED were 29 and 30% higher, and those of TA were 33% lower in the elderly as compared to the young, respectively. Average values during stance of GMAX, GMED and GMIN in the elderly were also 52, 60 and 40% higher than those of the young, respectively. In addition, average values during swing phase of GMED, HAM, GAS and SOL in the elderly were 22, 19, 25 and 28% lower than those of the young, respectively. Average value during stance phase of IL had significant interaction between age and gender, and only elderly men had significantly lower average value during stance phase of IL as compared to young men. Although the average value during stance phase of RF also had significant interaction between age and gender, there was no significant difference between the elderly and the young in both men and women.

Table 4-2 Peak values of joint moment.

Moment (N/kg)		Elderly		Young	
		Men	Women	Men	Women
Hip	Extension	0.71 (0.31)	0.58 (0.29)	0.69 (0.19)	0.52 (0.16)
	Flexion†	-0.70 (0.20)	-0.61 (0.19)	-0.73 (0.16)	-0.57 (0.12)
Knee	Flexion during early stance†	-0.19 (0.15)	-0.06 (0.16)	-0.18 (0.10)	-0.11 (0.09)
	Extension during early stance	0.62 (0.24)	0.81 (0.22)	0.76 (0.19)	0.75 (0.14)
	Flexion during terminal stance†	-0.25 (0.08)‡	-0.02 (0.16)*,‡	-0.16 (0.14)	-0.19 (0.12)
	Extension during terminal stance	0.20 (0.08)	0.27 (0.14)	0.19 (0.09)	0.15 (0.06)
	Flexion during swing phase	-0.23(0.04)	-0.19(0.05)	-0.28(0.08)	-0.20(0.32)
Ankle	Dorsiflexion	-0.25 (0.11)	-0.24 (0.07)	-0.29 (0.08)	-0.23 (0.15)
	Plantar flexion†	1.21 (0.29)*	1.00 (0.12)*	1.32 (0.12)	1.23 (0.11)

Value: mean (Standard deviation: SD)

* : Significant difference between age groups, $p < 0.05$

† : Significant difference between gender groups, $p < 0.05$

‡ : Age × gender interaction, $p < 0.05$

The main effect of number of synergies was significant ($F_{6, 252} = 454.52$, $p < 0.001$), and NMF identified five muscle modules, as VAF significantly increased to the fifth of synergies (**Figure 4-7**). There was no significant interaction between number of synergies and age ($F_{6, 252} = 0.61$, $p = 0.72$), number of synergies and gender ($F_{6, 252} = 0.31$, $p = 0.57$), and number of synergy, age and gender ($F_{6, 252} = 0.27$, $p = 0.94$). When the module profiles and weightings were used to reconstruct the muscle force data, they accounted for more than 90% of the muscle force variance for all muscles and regions in the gait cycle.

Muscle modules during walking are shown in **Figure 4-8**. Module 1 primarily contributes to weight acceptance in early stance. GMED, GMIN and VAS mainly act in module 1. Muscle weighting of TA significantly increased in the elderly. Moreover, muscle weighting of SOL had significant interaction between age and gender, and only elderly women had significantly lower muscle weighting as compared to young women. Module 2 contributes to propulsion in late stance. GAS and SOL mainly act in module 2. Muscle weightings of GMAX, RF and TA significantly increased in the elderly. Module 3 contributes to body support during stance phase. IL and RF mainly act in module 3. Muscle weighting of IL significantly decreased in the elderly. In addition, muscle weighting of GAS had significant interaction between age and gender, and only elderly women had significantly higher muscle weighting as compared to young women. Module 4 contributes to decelerate the leg motion and prepare for weight acceptance in

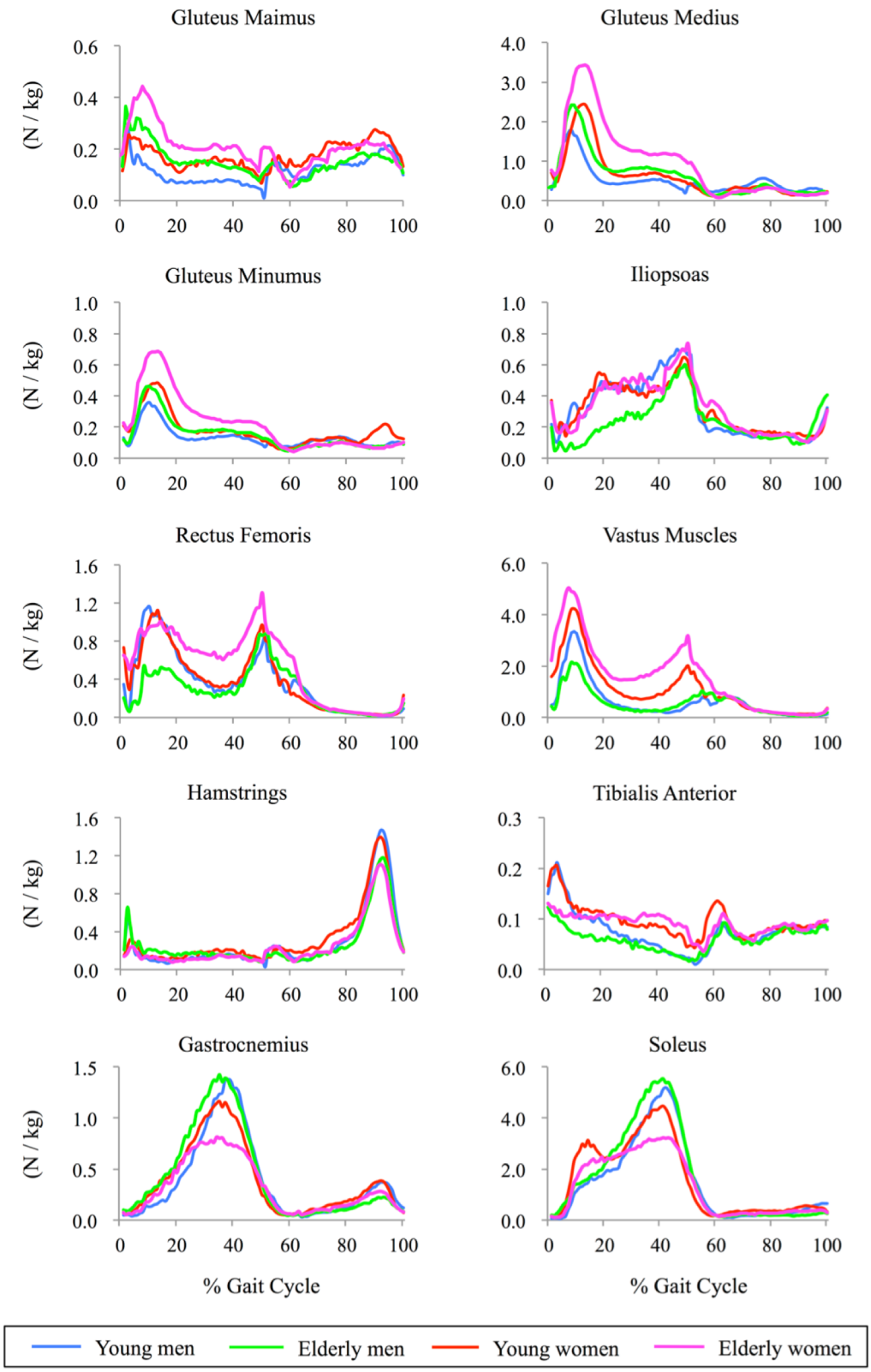


Figure 4-5 Muscle tension force during a gait cycle obtained by OpenSim.

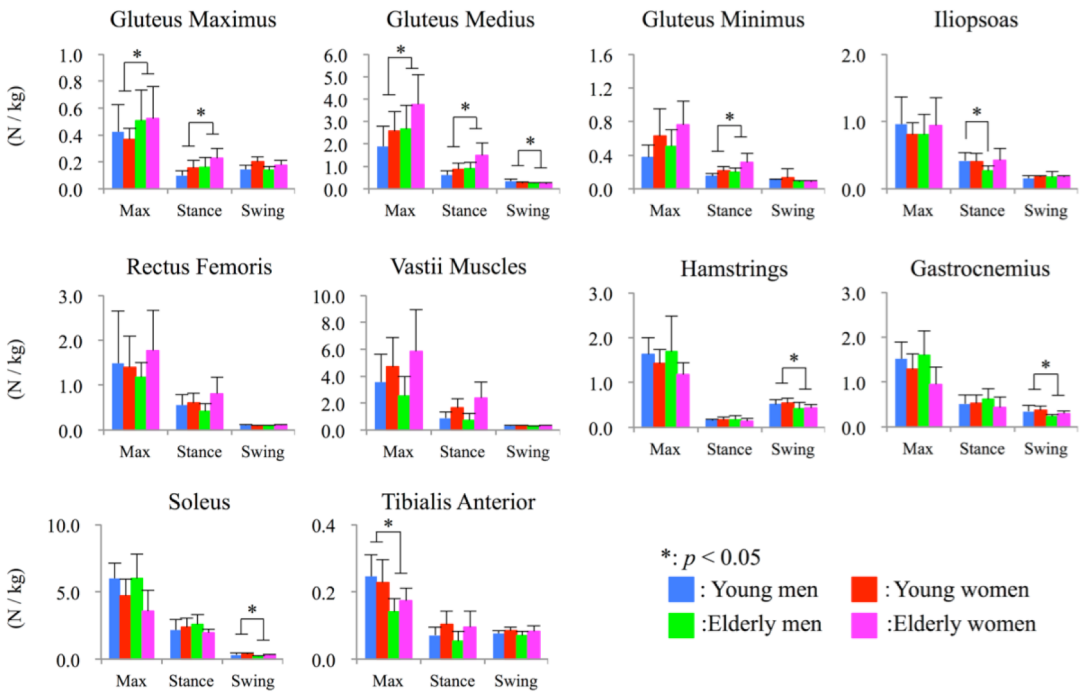


Figure 4-6 Comparison of the peak values during a gait cycle, and average values during stance and swing phase of each muscle force between the elderly and young people both men and women. Error bars span ± one standard deviation.

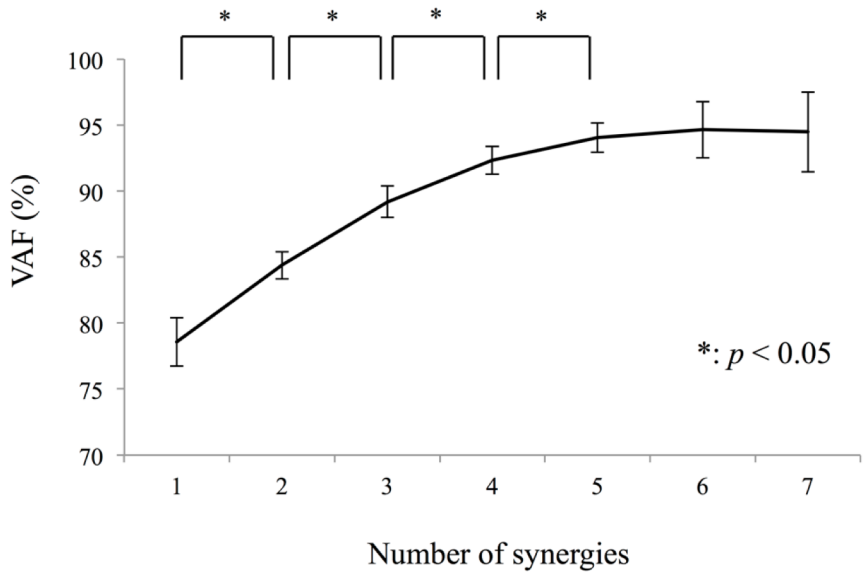


Figure 4-7 Variability accounted for (VAF) for determination of the number of synergies, averaged over the test subjects (n = 40). VAF indicate the ratio of consistency of both the shape and magnitude of the original and reconstructed data sets. Error bars span ± one standard deviation.

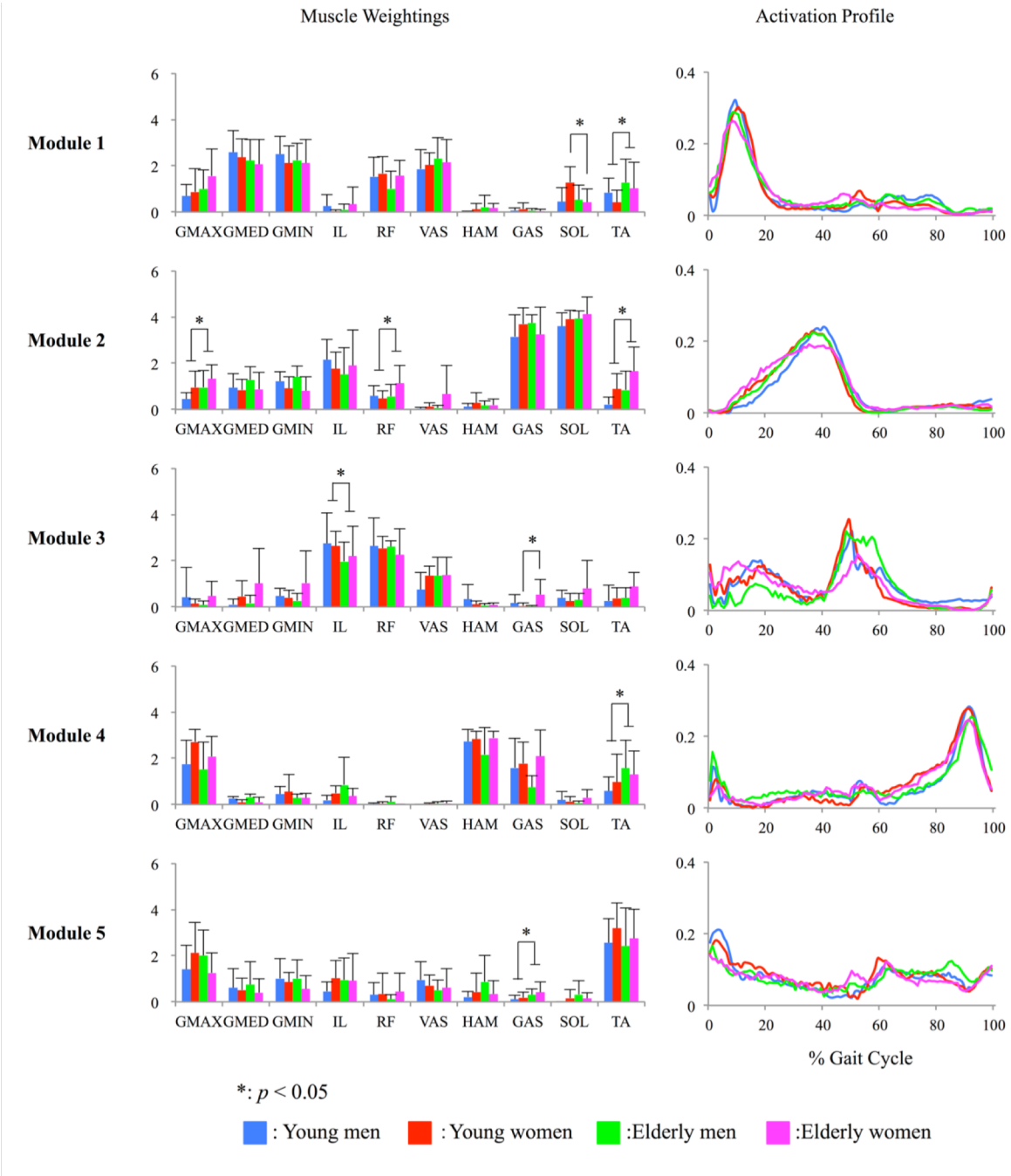


Figure 4-8 Muscle module weightings and activation timing profiles during walking derived from NMF analysis. Muscle weightings derived from NMF analysis indicate relative contribution of each muscle within the corresponding module. *$p < 0.05$ for the difference between the elderly and the young. Error bars span ± one standard deviation. Activation profiles indicate when the muscle module is active during the gait cycle. Each line represents the mean of each group.

Table 4-3 Similarities of activation pattern of muscle module between the elderly and the young.

	Module 1	Module 2	Module 3	Module 4	Module 5
Men	0.97*	0.97*	0.79*	0.94*	0.78*
Women	0.95*	0.97*	0.79*	0.97*	0.75*

Values are cross-correlation coefficient of activation profiles. Correlation coefficient (r) is interpreted as large ($r > 0.7$).

*: $p < 0.05$

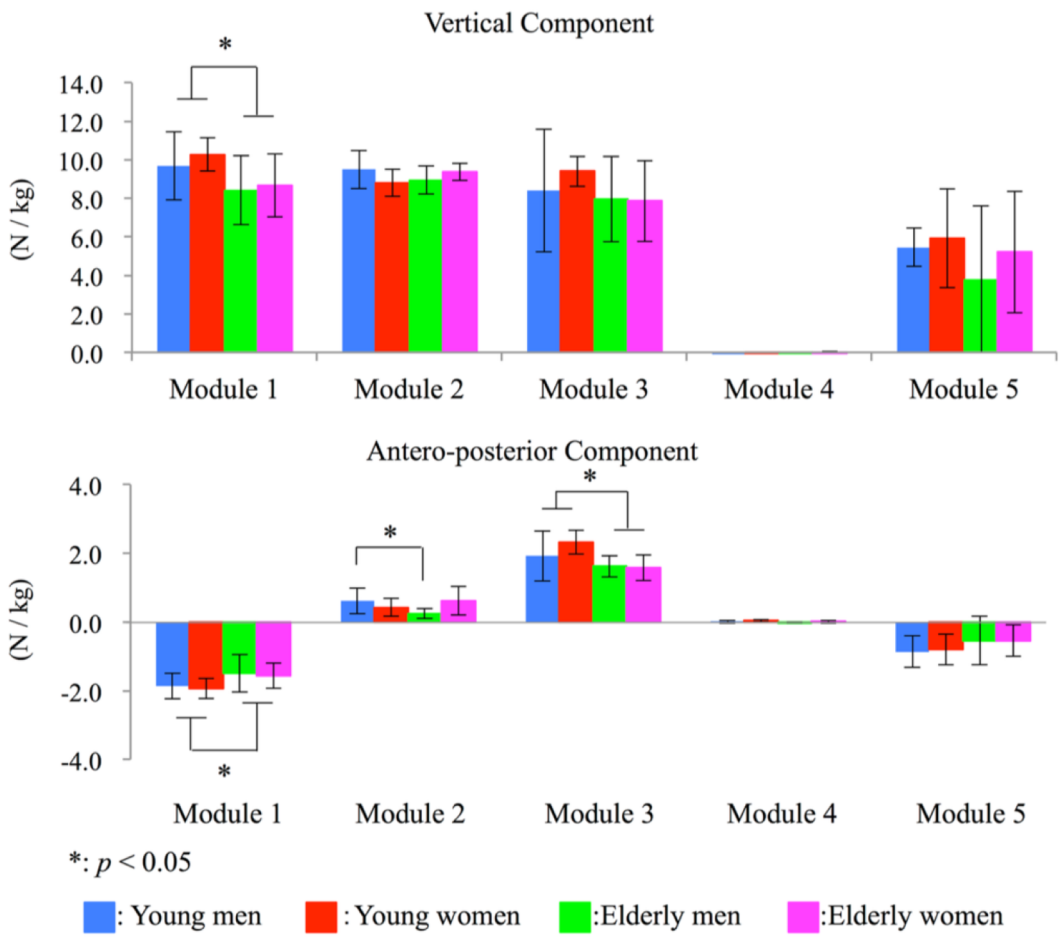


Figure 4-9 Vertical and antero-posterior ground reaction force profiles corresponding to each muscle module. Error bars span ± one standard deviation. *$p < 0.05$ for the difference between the elderly and the young.

late swing. GMAX and HAM mainly act in module 4. Muscle weighting of TA significantly increased in the elderly. Module 5 contributes to weight acceptances at initial contact. TA mainly acts in module 5. Muscle weighting of GAS significantly increased in the elderly although the contribution was small.

The cross-correlation coefficient between the elderly and young people is large ($r > 0.70$) in all modules, and the activation timing profiles of the elderly and the young were highly similar (**Table 4-3**).

The ground reaction force profiles corresponding to each muscle module are presented in **Figure 4-9**. Both vertical and antero-posterior components corresponding to module 1 were significantly lower in the elderly as compared to young people. Antero-posterior component corresponding to module 3 of the elderly was significantly lower than those of the young. In addition, antero-posterior component corresponding to module 2 were significantly lower in the elderly men than the value for young men.

4.4. Discussion

This study investigated the age-related differences of muscle control for support and propulsion during walking in both men and women. This study revealed that muscle activity of gluteus muscles of the elderly was excessive during stance phase. Moreover, there was an age-related change in the coordination of muscles around the ankle, and muscles of the lower extremity exhibited co-contraction in late stance in the elderly. In addition, these age-related alterations of muscle control might be associated with

support and propulsive forces during walking. These findings support our hypothesis that there were the age-related alterations of muscle coordination during walking, and the support and propulsive force corresponding to these muscle modules were reduced in the elderly.

In the elderly, the average values during stance phase of GMAX, GMED and GMIN were higher than the value for the young. Muscle activation during walking is generally higher in the elderly, although their classifications were based on observation of EMG data rather than quantitative evaluation (Finley et al., 1969). Elderly people have less hip extension angle during walking (Kerrigan et al., 1998a). The elderly exhibit muscle activity of gluteus muscles for the support in hip bending posture during stance phase. On the other hand, muscle activation during stance phase of IL in the elderly men was significantly lower than that of young men. Because only elderly woman had a significant relationship between the muscle cross-sectional area of the psoas major and walking ability (Kim et al., 2000), function of IL seems to be important for walking of elderly women. Moreover, they had a larger hip flexor moment during late stance, as compared to the men (Boyer, et al., 2008). Therefore, the function of IL in the elderly men is likely to decrease during walking. Muscle activations during swing phase of GMED, HAM, GAS and SOL in the elderly were significantly lower than those of the young, although the amplitude of activity of GMED, GAS and SOL were small. Activation of HAM is important to decelerate the leg in late swing. Therefore, the elderly might become unable to control the movement of the leg during swing phase. Elderly people had increased amount of muscle force in gluteus muscles, especially

during stance phase although the muscle strength of lower extremity has declined (Moreland et al., 2004). It may be necessary for the elderly to intervene not only to strengthen the weak muscle, but also to prevent excessive muscle activity.

The timing and composition of each module obtained in this study corresponded to a key phase of the gait cycle, consistent with the biomechanical subtasks of walking. In this study, five muscle modules were obtained in all groups. The activation pattern in each module was approximately consistent with previous studies (Ivanenko et al., 2004, 2005). Moreover, timing and shape of these modules were similar between elderly and young people in both men and women. Ivanenko et al. (2004) reported that there were common basic patterns, even when walking conditions were changed. The age-related change of the underlying activation timing profile might be also small. On the other hand, there were age-related alternations in the contribution of muscles to each module. The pattern and combination of muscle activation were regulated by interneurons in the spinal cord that are part of dispersed CPG (Goldings, 2009). Elderly people had changed combinations of muscle rather than the activation pattern in order to achieve the walking subtasks.

GMED, GMIN and VAS contributed to weight acceptance in early stance. These muscles generate vertical and backward accelerations of the center of gravity, and contribute to support and braking on the body (Liu et al., 2006). Moreover, contribution of TA significantly increased in the elderly. TA also generates vertical and backward accelerations of center of gravity in early stance (Liu et al., 2006). Elderly people have muscle weakness of lower extremity, especially in leg extensors (Moreland et al., 2004).

Consequently, muscles of the ankle joint as well as hip and knee joint also participate in weight acceptance. In addition, contribution of SOL was large only in young women, and significantly decreased in elderly women. SOL also generates only a small amount of vertical and backward accelerations of center of gravity in early stance (Liu et al., 2006), and may assist support and braking of the body in young women. Especially in women, there are age-related changes in the muscle coordination of the ankle in early stance. Both vertical and antero-posterior component corresponding to this module were significantly lower in the elderly as compared to the young. The age-related alteration of the ankle control in early stance may affect the support and braking force.

GAS and SOL contributed in late stance to generate vertical and forward accelerations of the center of gravity, and provide support and propulsion on the body (Liu et al., 2006). Moreover, both elderly men and women had significantly high muscle weightings of GMAX, RF and TA. This result suggested that many muscles in lower extremity in late stance were co-contracting in the elderly. Co-contraction of lower extremity muscles can be predictive of performance on a dynamic balance test (Nelson-Wong et al., 2012), and risk of falling during walking (Nagai et al., 2012) in the elderly. Co-contraction of lower extremity muscles in late stance in the elderly may compensate for low balance ability. Only elderly men had significantly lower anterior component of ground reaction force corresponding to this module compared to young men. Therefore, co-contraction in late stance may lead to low propulsive force with slow walking speed in elderly men. On the other hand, there was no difference in this force between the elderly and the young women. Elderly women may generate

propulsive force using various muscles, and there was a gender difference in the effect of co-contraction in late stance on propulsive force.

RF and IL contributed to support, braking and propulsion of body in early and late stance. Neptune et al. (2008) found that the hip flexors, IL and RF made larger contributions to swing initiation and trunk propulsion, respectively. Muscle weighting of IL in the elderly significantly decreased for this module, and antero-posterior component of the ground reaction force corresponding to this module of the elderly was significantly lower than the value for the young. Therefore, functional decline of IL in the elderly affect the propulsion in late stance, and the elderly depended on the RF in performance of the stance-to-swing transition. Moreover, only elderly women had large muscle weighting of GAS. Support, braking and propulsion due to this module is performed by co-contraction between the hip flexor and the ankle plantar flexor in the elderly women.

GMAX and HAM mainly act in late swing in module 4. The ground reaction force corresponding this module was very small, and this module was not directly involved in the generation of ground reaction force. HAM is activated to decelerate the leg during swing phase in preparation for foot contact (Liu et al., 2006). Moreover, both elderly men and women had significantly high muscle weightings of TA. TA generates vertical and backward accelerations of the center of gravity in early stance (Liu et al., 2006). TA in module 1 of the elderly also had significantly high muscle weightings. It follows that TA of the elderly greatly contributed to preparation of foot contact from late stance to early stance.

TA contributes to weight acceptance at initial contact through co-activation of GMAX. These muscles have the role of shock absorption at initial contact. The age-related alteration of this module did not affect the ground reaction force. The muscle weighting of GAS of the elderly was significantly greater. The ankle of the elderly accepted their body weight by co-contraction between dorsiflexor and plantar flexor at initial contact.

Weighting of muscles in the ankle was significantly different between the elderly and the young in all modules. There was an age-related change in the coordination of the ankle. De Vita and Hortobagyi (2000) reported that the elderly showed a redistribution of joint moments and powers, which emphasized proximal extensors and de-emphasized distal extensors compared with the young. Judge et al. (1996) reported that peak ankle joint power generation at pre-swing decreased in the elderly. It is necessary for improvement of elderly's ability to walk to intervene the ankle control in addition to muscle strengthening, because the influence of functional decline with age is readily apparent in the ankle joint during walking.

In this study, the analysis of muscle tension force was performed using musculoskeletal simulation. Generally, surface electromyography (sEMG) is often used for the analysis of muscle activity during motion. However, the results of sEMG are affected by not only the amount of muscle activity, but also the thickness and the conductivity of the skin and subcutaneous fat. Basmajian et al. (1985) reported that the integrated value of muscle activity potentials and the muscle force during isometric contraction has linear or curvilinear relationship. On the other hand, the muscle tension

force determined by musculoskeletal simulation is calculated from the kinematics data, kinetic data, and physical parameters obtained by the three-dimensional motion analysis system, and with the exception of the uncertainties as premises for the used model. As the estimated muscle tension forces were analyzed, it is necessary to confirm the reliability of the estimated data. The waveform of the muscle force used in this study is similar to the waveform of the muscle activity of the lower extremity reported by Winter (1991). Therefore, the reliability of the estimated data of the muscle tension force used in this study was assumed to be good.

4.5 Summary

In this chapter, the objectives were to reveal the age-related differences of the amplitude, timing and combination of muscle activation during walking, and the relationship between muscle modules and support and propulsive force during walking. Twenty community-dwelling elderly people and 20 younger people participated this study.

The results are as follows: 1) muscle activity of gluteus muscles of the elderly was excessive during stance phase, 2) there was an age-related change in the coordination of muscles around the ankle, and muscles of the lower extremity exhibited co-contraction in late stance in the elderly, 3) these age-related alterations of muscle control might be associated with support and propulsive forces during walking.

In conclusion, muscle control during walking is age-related change in the qualitative and quantitative. Since the elderly increased muscle activity of gluteus

muscles, and the number of muscles that contribute in each subtask, it was necessary to inhibit activity of the unneeded muscle. In addition, there are gender differences in the age-related differences in muscle control of the lower extremity.

Chapter 5

Conclusion

This dissertation addressed the question about the age-related alteration of walking control. The primary aim of this study was to find the differences in the control of the joint moment and muscle of the lower extremity for support and propulsion. The differences in the effect of walking speed on the stride-to-stride acceleration variability between the elderly and the young were also investigated. The present study revealed the age-related differences in control strategy of walking from various aspects. It will be extended such investigation to further access walking control mechanism, and developed the effective treatment methods and equipment for improvement the function of central nervous system of the elderly.

5.1 Conclusion

The results obtained in this dissertation could be summarized as follows:

Chapter 1 presented background and the motivation for this dissertation, defined its objective and exhibited the overview of this dissertation. Regarding the literature reviews, the previous researches perspective for the age-related alterations of

the spatiotemporal parameters, walking variability, kinematic data, kinetic data and muscle activity were focused.

Chapter 2 investigated the stride-to-stride acceleration variability through simultaneous evaluation of the acceleration of head, lumbar, and lower extremity as a function of walking speed, and clarified whether the effect of walking speed on this acceleration variability was different between the body parts in the elderly and the young. The effects of walking speed on the stride-to-stride acceleration variability of all body parts of the elderly and the young were same trend, and that of the lower extremity was affected when walking slowly in both the elderly and the young in AP and ML directions. Especially in ML direction, there is laterality in the effect of walking speed, and the acceleration variability on the left lower extremity was less affected by walking speed. In vertical direction, acceleration variability was strictly controlled so as to stabilize the walking behavior despite the effect of walking speed. Additionally, the elderly was too stable acceleration pattern, regardless of changing walking speed.

Chapter 3 investigated the differences in the relationships between the vertical and antero-posterior ground reaction forces, and the lower extremity joint moment during walking between the elderly and the young. In young men, extension moment during early and late stance of the knee related to the minimum value at mid-stance, second peak force of the vertical ground reaction force and the anterior component of the ground reaction force, respectively. Extension moment of the hip related to the posterior component of the ground reaction force. In elderly men, extension moment during early stance of the knee related to the posterior component of the ground reaction

force, although there were no factors related to the vertical ground reaction force. In young women, the knee joint moment related to the first peak force, and the minimum value at mid-stance of the vertical ground reaction force, and antero-posterior ground reaction force. In the elderly women, all of the hip, knee and ankle joint moments were related to the vertical and antero-posterior ground reaction forces. Therefore, the elderly and the young used the lower limb joints differently for the control of the vertical and antero-posterior ground reaction force.

Chapter 4 investigated that the age-related differences of amplitude, timing and combination of muscle activation during walking, and the relationship between muscle modules and support and propulsive force during walking. Muscle activity of gluteus muscles of the elderly was excessive during stance phase, and there was an age-related change in the coordination of muscles around the ankle, and muscles of the lower extremity exhibited co-contraction in late stance in the elderly. In addition, these age-related alterations of muscle control might be associated with support and propulsive forces during walking. Therefore, results of this study suggested that muscle control during walking is age-related change in the qualitative and quantitative.

In this dissertation, the age-related alterations of walking control were studied from various aspect. Walking variability as a function of walking speed was discussed in Chapter 2. The control of joint moment and muscle of lower extremity for support and propulsion of walking was discussed in Chapter 3 and Chapter 4, respectively. The elderly had age-related alteration in control of the lower extremity for support and propulsion during a gait cycle. Additionally, muscles of the lower extremity of the

elderly exhibited co-contraction during walking, and the elderly performed more rigid or stiff walking control than the young, regardless of changing walking speed. These study results revealed that control strategy of walking was changed with age, and the elderly had qualitative change in neuro-muscular function of walking by central nerve system. Therefore, muscle strength exercise by isolated movement, sensorimotor integration for muscle control, and phase-dependent muscle coordination exercise using EMG biofeedback in order to inhibit co-contraction of the lower extremity are necessary for walking control of the elderly.

5.2 Future work

This section shows future directions on age-related alterations of walking control and provides general discussions based on our research findings.

1) Development a model and simulation system incorporated the physical features of a subject

The model used in Chapter 4 was identical between the young and the elderly. The influence of muscle mass difference between the young and the elderly can be removed by normalization according to the subject's body weight, and the dimensions of the segments can be scaled to match the subject in OpenSim. However, Gajdosik (2005) reported that the elderly women had greater percent increases for the average passive torque during passively moving the ankle from plantar flexion into dorsiflexion than the

young women, and calf muscle-tendon unit of the elderly women had increased passive viscoelastic properties. Additionally, alignment of spine and pelvis changes with aging (Hammerberg, 2003), although subject in this study did not have structural deformation of spine column. Viscoelastic change in muscles and the deformation of spine could not be incorporated into the model. In the future, in order to perform musculoskeletal simulation with higher accuracy, it is necessary to develop a model and simulation system incorporating the physical features of a subject, such as age, gender and joint deformity.

2) Specific of factors affecting the age-related alterations of walking control

There were age-related alterations of the control of acceleration variability from Chapter 2, and the muscle control during walking from Chapter 4. Muscle force of gluteus muscles of the elderly was large during stance phase despite they had muscle weakness. Elderly people have a functional decline of the musculoskeletal system and nervous system with age. Further studies are required to determine the relationship between walking control and physical function in the elderly in order to develop exercise regimen for the elderly.

3) Clinical application of results of this dissertation

Previous study recommended muscle strength training for reduction of walking ability of the elderly (Oesen et al., 2015). However, this study revealed that walking control of the elderly differs from that of the young. These results suggested that it was necessary

to intervene for alteration output balance between joint and coordination exercise for spatial component that is combination of muscles and the timing component that is reaction time in addition to strength training for amplitude component of muscle. Problems for reduction of walking ability of elderly people include age-related changes of motor control by central nervous system as well as musculoskeletal system.

In the future, it is necessary to develop effective treatment methods and equipment for improvement of function of central nervous system. Moreover, further research focused on walking control is useful to establish evidence for maintaining the walking ability of the elderly, and lead to extension of healthy life expectancy.

Moreover, there were gender differences in age-related alterations in walking control as well as body structure. It is also necessary to develop gender-specific treatment methods for maintaining the ability to walk.

Appendix A
Calculation methods of joint moment

A free-body diagram of each segment for calculation of joint moments in sagittal plane is shown in **Figure A-1**. Muscles produce moments of force across joints during walking (Soft tissues, like ligaments, and skin resistance, can also produce forces and moments when they are elongated).

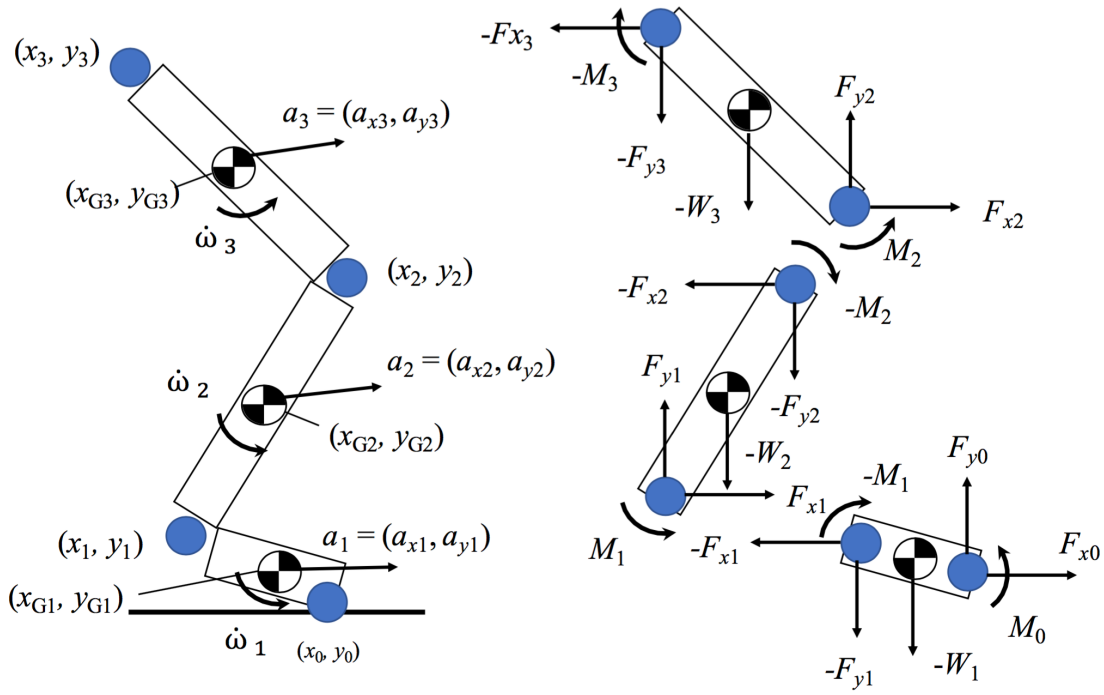


Figure A-1 A free body diagram of supporting leg during walking.

The rotational equations of motion in foot segment are:

$$F_{x1} = F_{x0} - m_1 a_{x1}$$

$$F_{y1} = F_{y0} - m_1 a_{y1} - W_1$$

$$M_1 = (x_0 - x_{G1})F_{y0} - (y_0 - y_{G1})F_{x0} - (x_1 - x_{G1})F_{y1} + (y_1 - y_{G1})F_{x1} - I_1\dot{\omega}_1$$

The rotational equations of motion in shank segment are:

$$F_{x2} = F_{x1} - m_2 a_{x2}$$

$$F_{y2} = F_{y1} - m_2 a_{y2} - W_2$$

$$M_2 = M_1 + (x_1 - x_{G2})F_{y1} - (y_1 - y_{G2})F_{x1} - (x_2 - x_{G2})F_{y2} + (y_2 - y_{G2})F_{x2} - I_2\dot{\omega}_2$$

The rotational equations of motion in thigh segment are:

$$F_{x3} = F_{x2} - m_3 a_{x3}$$

$$F_{y3} = F_{y2} - m_3 a_{y3} - W_3$$

$$M_3 = M_2 + (x_2 - x_{G3})F_{y2} - (y_2 - y_{G3})F_{x2} - (x_3 - x_{G3})F_{y3} + (y_3 - y_{G3})F_{x3} - I_3\dot{\omega}_3$$

where I_1, I_2, I_3 = moments of inertia of each segment

m_1, m_2, m_3 = mass of each segment

x_1, x_2, x_3, y_1, y_2, y_3 = coordinate of joint center of each joint

x_{G1}, x_{G2}, x_{G3}, y_{G1}, y_{G2}, y_{G3} = coordinate of center of mass of each segment

$\dot{\omega}_1, \dot{\omega}_2, \dot{\omega}_3$ = angular velocity of each segment

a_1, a_2, a_3 = angular acceleration of each segment

W_1, W_2, W_3 = gravitational force of each segment

M_1, M_2, M_3 = previously calculated distal moments of each segment

F_0 = ground reaction force

F_1, F_2, F_3 = previously calculated joint reaction force of each segment

Body segment inertia parameters (Okada et al., 1996) used to calculate of joint moments are shown as follow:

Table A-1 Body segment inertia parameters of the elderly and the young.

	Eldery men					Young men				
	Mass	CG	RG			Mass	CG	RG		
			kx	*ky*	*kz*			*kx*	*ky*	*kz*
Trunk	0.497	0.498	0.368	0.372	0.199	0.489	0.493	0.346	0.357	0.167
Thigh	0.092	0.481	0.272	0.283	0.171	0.110	0.457	0.278	0.270	0.152
Shank	0.047	0.423	0.263	0.284	0.144	0.051	0.406	0.274	0.271	0.097
Foot	0.017	0.581	0.233	0.126	0.235	0.011	0.595	0.177	0.088	0.182
	Eldery women					**Young women**				
	Mass	CG	RG			Mass	CG	RG		
			kx	*ky*	*kz*			*kx*	*ky*	*kz*
Trunk	0.493	0.515	0.368	0.370	0.214	0.457	0.506	0.343	0.355	0.170
Thigh	0.098	0.474	0.277	0.289	0.181	0.123	0.458	0.285	0.278	0.157
Shank	0.048	0.424	0.277	0.289	0.181	0.053	0.410	0.275	0.272	0.102
Foot	0.015	0.591	0.226	0.122	0.229	0.011	0.594	0.184	0.090	0.189

Mass, CG, RG is indicated as segment mass, location of segment center of gravity, segment radius of gyration, respectively.

Mass is given as the ratio to total body mass.

CG is given as the ratio to each segment length measured from the proximal end of the segment.

RG is given as the ratio to each segment length around *x-y-z* axes.

Appendix B
Musculotendinous model

OpenSim supplies multiple muscle models based on formulations of muscle dynamic simulation of movement (Zajac, 1989). The muscle contractile dynamics is represented as a Hill-type contractile element relating muscle force to activation, muscle-fiber length and contractile velocity are generalized to a model of whole muscle contraction dynamics (**Figure B-1**) (Seth, et al., 2011).

$$F^{MT} = F^{T} = F^{M}\cos(\alpha)$$

$$F^{T} = k^{T}(l^{MT} - l^{M}\cos(\alpha) - l^{ST})$$

$$F^{M} = a \cdot f_{l}(l^{M}) f_{v}(V^{M}) + f_{PE}(l^{M})$$

$$\dot{L}^{M} = f_{v}^{-1}\left[\frac{k^{T}(l^{MT} - l^{M}\,cos(\alpha) - l^{ST})/cos\,(\alpha) - f_{PE}(l^{M})}{a \cdot f_{l}(l^{M})}\right]$$

where F^M and F^T are muscle and tendon force, a is muscle activation, L^M is muscle-fiber length, k^T and L^{ST} are the tendon stiffness and the slack length properties, and f_l and f_v are the physiological force-length and force-velocity relationships of the contractile element and f_{PE} is the passive-element force-length curve for inputs that are not normalized. The equations of the force-velocity and force-length relationships indicated by Hill are indicated as follows:

$$(v + b)(F^M + a) = b(F_0^M + \alpha)$$

$$f_l(l^M) = 1 - \frac{(L^M - L_0^M)^2}{L_{width}^2}$$

where v is the velocity of contraction, α and b are constant force and constant velocity, F_0^M is the maximum isometric muscle force, and L_{width} is is the spread of muscle. Muscle-tendon parameters for lower extremity muscles used in the simulation into OpenSim were indicated in **Table B-1**.

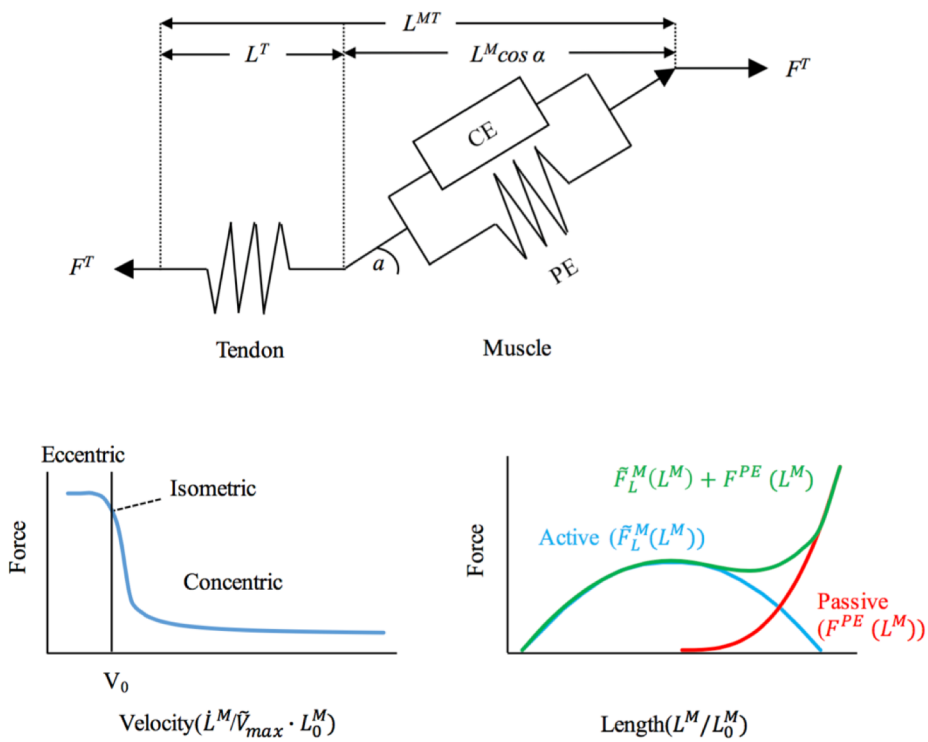


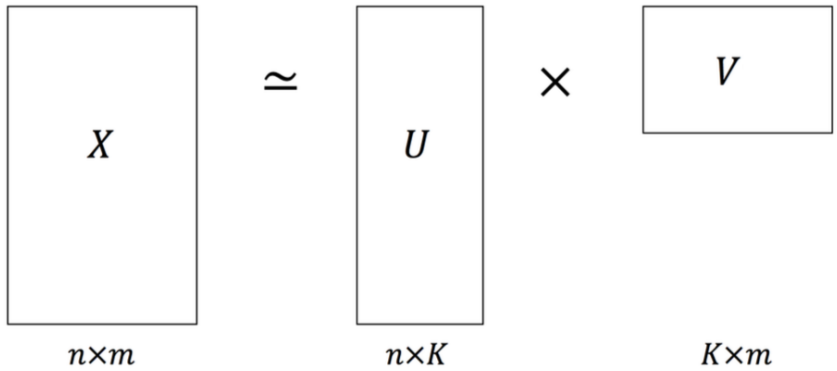
Figure B-1 Schema representation of the elements of a musculotendinous actuator as a single muscle in series with an elastic tendon. *F* is the scholar force or tension, and *L* is length with superscripts referring to muscle *M*, tendon *T* or the complete musculotendinous actuator *MT*. The pennation angle α represents the mean orientation of muscle fibers in a muscle and changes as function of L^M. Graphical representation of the force generating capacity of contractile element (*CE*) and parallel elastic element (*PE*) of the muscle as a function of normalized shortening velocity and muscle-fiber length.

Table B-1 Muscle-tendon parameters for lower extremity muscles (Delp, 1990).

muscle	peak muscle force (N)	optimal fiber length (cm)	pennation angle (α)	tendon slack length (cm)	tendon length /fiber length
hip muscles					
gluteus medius1	550	5.4	8	7.8	1.4
gluteus medius2	380	8.4	0	5.3	0.6
gluteus medius3	435	6.5	19	5.3	0.8
gluteus minimus1	180	6.8	10	1.6	0.2
gluteus minimus2	190	5.6	0	2.6	0.5
gluteus minimus3	215	3.8	1	5.1	1.3
gluteus maximus1	380	14.2	5	12.5	0.9
gluteus maximus2	550	14.7	0	12.7	0.9
gluteus maximus3	370	14.4	5	14.5	1.0
adductor magnus1	345	8.7	5	6.0	0.7
adductor magnus2	310	12.1	3	13.0	1.0
adductor magnus3	445	13.1	5	26.0	2.0
adductor longus	420	13.8	6	11.0	0.8
adductor brevis	285	13.3	0	2.0	0.2
pectineus	175	13.3	0	0.1	0.1
iliacus	430	10.0	7	9.0	0.9
psoas	370	10.4	8	13.0	1.3
quadratus femoris	255	5.4	0	2.4	0.4
pemelli	110	2.4	0	3.9	1.6
piriformis	295	2.6	10	11.5	4.4
hip and knee muscles					
rectus femoris	780	8.4	5	34.6	4.0
semimembranosus	1030	8.0	15	35.9	4.5
semitendinosus	330	20.1	5	26.2	1.3
biceps femoris (lh)	720	10.9	0	34.1	3.1
gracilis	110	35.2	3	14.0	0.4
sartorius	105	57.9	0	4.0	0.1
tensor fasciae latae	155	9.5	3	42.5	4.5
knee muscles					
vastus medialus	1295	8.9	5	34.6	4.0
vastus intermedius	1235	8.7	3	13.6	1.6
vastus lateralis	1870	8.4	5	15.7	1.9
biceps femoris (sh)	400	17.3	23	10.0	0.6
knee and ankle muscles					
medial gastrocnemius	1115	4.5	17	40.8	9.0
lateral gastrocnemius	490	6.4	8	38.5	6.0
ankle muscles					
soleus	2830	3.0	25	26.8	8.9
tibialis posterior	1270	3.1	12	31.0	10.0
fl. digitorum longus	310	3.4	7	40.0	11.8
fl. hallucis longus	320	4.3	10	38.0	8.8
peroneus brevis	350	5.0	5	16.1	3.2
peroneus longus	755	4.9	10	34.5	7.0
tibialis anterior	600	9.8	5	22.3	2.2
peroneus tertius	90	7.9	13	10.0	1.3
ext. digitorum longus	340	10.2	8	34.5	3.4
ext. hallucis longus	110	11.1	6	30.5	2.6

Appendix C
Non-negative matrix factorization

Non-negative matrix factorization (NMF) analysis is linear decomposition technique that assumes that the set of measured data is composed of linear combinations of a smaller number of underlying elements (Ting and Chvatal, 2010). Potential elements of the original matrix can be clearly show by NMF. If matrix X that is non-negative value is indicated $n \times m$, and the number of clusters of interest is K, X is separated into two matrixes U and V in NMF:



$$X \simeq UV$$

Figure C-1 Decomposition of a matrix by NMF.

where matrix U is $n \times K$, and matrix V is $K \times m$. Each component is a non-negative value.

In decomposition, the sum of the squares of the matrix in each component (Frobenius norm) was used as the cost function.

$$\|A - B\|^2 = \sum_{ij} \left(A_{ij} - B_{ij}\right)^2$$

Matrix U and V are calculated from non-negative matrix X, and the number of clusters K by repeatedly using the following formulas:

$$V'_{ij} \leftarrow V_{ij} \frac{(X^t U)_{ij}}{(V U^t U)_{ij}}$$

$$U'_{ij} \leftarrow U_{ij} \frac{(X^t V)_{ij}}{(U V^t V)_{ij}}$$

where U_{ij} and V_{ij} are show the elements of the i-th row and j-th column in U and V. U'_{ij} and V'_{ij} are obtained from U and V by using the above formulas, respectively. U' and V' are regarded as new U and V, and the above formulas are repeated. Because the values of $\|X - UV\|^2$ is gradually reduced by the repetition of the above formulas, U and V that the errors of decomposition are minimum are able to be calculated. However, the values of $\|X - UV\|^2$ may not converge to the true minimum values. The initial values $U^{(0)}$ and $V^{(0)}$ used the above formulas influence the results of U and V. Therefore, U and V obtained by repeatedly using the above formulas are low rank approximation.

Publications

This dissertation is based on the research that was previously reported in the following publications.

Journals

Toda, H., Nagano, A., Luo, Z. (2015). Age and gender difference in the control of vertical ground reaction force by hip, knee and ankle joints. *Journal of Physical Therapy Science* 27 (6), 1833-1838.

Toda, H., Nagano, A., Luo, Z. (2016). Gender difference in muscle force of lower extremity during walking in the healthy youths [in Japanese]. *Rigakuryouhoukagaku* 31 (1), 37-42.

Toda, H., Nagano, A., Luo, Z. (2016). Age-related differences in muscle control of lower extremity for support and propulsion during walking. *Journal of Physical Therapy Science* 28 (3), 794-801.

Toda, H., Nagano, A., Luo, Z. (2016). Analysis of walking variability through simultaneous evaluation of the head, lumbar, and lower extremity acceleration in the healthy youth. *Journal of Physical Therapy Science* 28 (6), 1801-1805.

International Conference

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