NAGOYA UNIVERSITY

Graduate School of Engineering

Gait analysis of the reaction motions against muscle restriction for the purpose of reproducing the gait of the elderly in the young

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Abstract

The overall aim of this doctoral project is the investigation and validation of Muscle Activity Restriction Taping Technique (MARTT), a technique proposed by the author of this project, that could serve to ease the understanding and the study of age-related changes in the elderly gait by utilizing young individuals to take the place of the elderly with the main objective of preventing the exposure of the older adults to exhausting and dangerous experimental conditions that the realization of such studies might imply. In addition, this project aims to investigate how the human body compensates muscle weakness during walking and for what purpose.

Chapter 1 introduces the social background of this project and the importance of assisting the elderly gait in order to maintain in this population a healthy mobility level that is necessary to guarantee a good quality of life. In order to assist the elderly gait, the understanding of their gait impairments under different activities and terrain conditions is essential. This chapter includes the state of the art on the elderly age-related gait impairments and joint compensation motions that are the result of the gait alterations in the elderly. This chapter also describes existing age simulation suits that constitute a method for young individuals to feel what it is to be in the body of an older person and to get conscious about the elderly needs.

As described in Chapter 1, the elderly gait encompasses several disorders, including a lower minimum toe clearance (MTC) to the ground, which is a potential cause of tripping and falling while walking. Devices that assist in maintaining a safe MTC while the elderly walk could reduce such risks. However, the testing and development processes of such devices require experimental trials in conditions that imply high injury risks for the older population. Thus, the participation of the elderly in these processes to find effective assistance methods may jeopardize their safety. To avoid this issue, young individuals could substitute the elderly in the initial experimental process. In this regard, the author of this doctoral project proposed MARTT as a method to reproduce the lower MTC of the elderly in young adults, by applying this technique to the lower-limb to cause a muscle

weakening effect.

Chapter 2 covers the validation of MARTT as a technique that is able to reduce the MTC of young adults to that of the elderly. In total 10 male subjects participated in this study, and the walking trials were carried out on a treadmill. By the means of MARTT, two different muscle restriction approaches were studied at two different walking speeds. One approach restricted muscles at the shank and the other restricted simultaneously muscles at the shank and thigh. One walking speed corresponded to 3.5 km/h, considered as the average speed for young adults, and the second walking speed corresponded to 4 km/h, regarded as the average speed for older adults. In the two mentioned approaches and walking speeds, the MTC of the young subjects was reduced to a median value lower than 10.1 mm, which is within the range of the MTC values reported for the elderly in literature, that is about 7 – 13 mm. The reduction of the MTC significantly increased the foot-ground contacts during the swing phase of the young subjects. Foot-ground contacts constitute a major cause of tripping and falling in older adults. The foot-ground contact frequency was more than twice as that in normal walking (natural walking of the subjects without the restriction applied by MARTT) when the shank muscles were restricted, and more than five times when both the shank and thigh muscles were restricted. In addition, MARTT reproduced in the young subjects spatio-temporal parameters that characterize the elderly gait, which are: reduced period of the single support phase and shorter step length.

Moreover, the loss of muscle mass with aging and the consequent muscle weakness results in compensatory body motions during walking. Although these compensatory motions increase the cost of walking, they appear to be an attempt by the elderly to maintain a safe ambulation, as suggested in studies about the elderly gait. However, the relationship between the affected muscles, that perceive a certain level of weakness, and compensatory body motions along the gait cycle needs elucidation. Similarly, the purpose of the compensatory motions and the gait characteristics during walking that the human body prioritizes to compensate need to be investigated.

Chapter 3 describes the examination of the gait compensation strategies of the young subjects when walking with MARTT. The lower-limb showed an active kinematic compensation chain, in which joints that perceived no or less restriction compensated for the most compromised joint to prevent foot drop, knee hyperextension in the terminal stance phase, and knee hyperflexion in the loading response phase, and to maintain the step length. In addition, joints could compensate for themselves when the muscles acting on the other joints were unable to assist, as observed on the ankle joint that compensated for itself to prevent foot drop when the knee and hip flexor muscles were restricted. The observed compensation strategies agreed with a previously reported computational simulation about gait compensations appearing as consequence of muscle weakness. To add, similarities with the compensation strategies reported for the elderly were found.

Chapter 4 encompasses a discussion about the application of MARTT in the study of the elderly gait and how this technique could be useful to find out effective gait assistive methods and test assistive devices in the young until the control strategies of such devices are mature and safe enough to be tested in the elderly. This chapter also discusses about the possibility of combining MARTT with existing age simulation suits to reproduce in the young not only the lower MTC and joint compensation strategies seen in the elderly as a result of muscle weakness, but also other afflictions, such as sensory impairments and the reduced joint range of motion. Additionally, this chapter includes a discussion on the limitations of MARTT in relation to the reproduction of the elderly gait characteristics in the young, and also a discussion on the limitations and justifications of the experimental trials that have been conducted for the purpose of this doctoral project.

Chapter 5 concludes this doctoral thesis highlighting the main findings on the study about the validation of MARTT, that is the focus of Chapter 2, and on the study about the biomechanical analysis of gait compensation strategies, that is the focus of Chapter 3. In addition, the future directions of this project, proposed by the author of this thesis, are listed in this chapter.

To conclude, this doctoral project has validated MARTT and has studied the compensation motions at the lower limb that result from muscle weakness. The results of this project provide insights on how to implement MARTT, its applicability and limitations, and also insights on joint compensations that appear with the deterioration of gait ability. Further studies should include the examination of the compensation strategies and classification proposed in this project in a larger number of young subjects, the analysis of the compensations at the upper body, and the comparison of electromyography data of the young walking with MARTT with that of the elderly.

List of Publications

The following lists the peer-reviewed journal articles that are contained in this thesis:

- J.B. Ullauri, Y. Akiyama, S. Okamoto and Y. Yamada, ”Technique to reduce the minimum toe clearance of young adults during walking to simulate the risk of tripping of the elderly,” *PLoS One*, vol.14, no.6, 2019

  doi.org/10.1371/journal.pone.0217336

- J.B. Ullauri, Y. Akiyama, S. Okamoto and Y. Yamada, ”Biomechanical analysis of gait compensation strategies as a result of muscle restriction,” *App. Sci.*, vol.11, no.18, 2021

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The following is the proceedings article where MARTT was introduced:

- J.B. Ullauri, Y. Akiyama, N. Yamada, S. Okamoto and Y. Yamada, ”Muscle activity restriction taping technique simulates the reduction in foot-ground clearance in the elderly,” *Proceedings of the 14th IEEE International Conference on Rehabilitation Robotics (ICORR)*, August 11-14, Singapore, pp. 559-564, 2015

  doi.org/10.1109/ICORR.2015.7281259

Other of my publications:

- J.B. Ullauri, Y. Akiyama, N. Yamada, S. Okamoto and Y. Yamada, ”Taping Technique to simulate the reduced minimum toe clearance of the elderly,” *Proceedings of the 2015 Annual Conference of the Robotics Society of Japan (RSJ)*, Tokyo, AC2L1-02, 2015.

- J.B. Ullauri, L. Peternel, B. Ugurlu, Y. Yamada, J. Morimoto, ”On the EMG-based torque estimation for humans coupled with a force-controlled elbow exoskeleton,” *Proceedings of the 2015 IEEE International Conference on Advanced Robotics (ICAR)*, July 27-31, Turkey, 6 pages, 2015

  doi.org/10.1109/ICAR.2015.7251472

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I would like to dedicate this work to my celestial father, for providing me always with the strength to reach my dreams no matter the difficulties that may appear on the journey.

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I would like to thank also my family Vitja, Nozomi and Feliz for their support and love, and for being there to motivate me in times when managing both my studies and my job was challenging and exhausting.

To finalize, I would like to express my love for Japan, the country that warmly welcomed me, and where many of my dreams came true.

Gaby

Statement of Originality

I hereby declare that this doctoral thesis is my own work, and that it has not been previously submitted for another degree or diploma in this or another university. Any source of information used in this work has been properly acknowledged.

Abbreviations

ADL	Activities of daily living
AGNES	Age Gain Now Empathy System (aging suit)
BFS	Biceps femoris short head
C-restriction	Calf restriction applied with MARTT
CT-restriction	Calf and thigh restriction applied with MARTT
FSR	Force sensing resistor
GAS	Gastrocnemius
GERT	GERonTologic aging suit
HAM	Hamstring
ILPS	Iliopsoas
IQR	Inter-quartile range
MARTT	Muscle Activity Restriction Taping Technique
MTC	Minimum toe clearance
NIA	National Institute on Aging
RF	Rectus femoris
ROM	Range of motion
SO	Soleus
TA	Tibialis anterior
VAS	Vastii
WHO	World Health Organization

Contents

Chapter 1

Introduction

1.1 Social background

1.1.1 Prevalence of elderly people in the world

For several years the world has been in a situation where the amount of older people and their life length have been increasing. According to a report on global health and aging [1] from the World Health Organization (WHO) and as observed in Figure 1.1, the world is already in a time where it is predicted that individuals above 65 years old outnumber children under 5 years old. This increment in the older population in comparison to the young is not only due to the longer life expectancy of the older, but also because of the lower fertility rates and the decision to have less children. Thus, with fewer new-born children and higher amount of people living longer, older individuals will constitute a bigger section of the world total population. It is projected that people aged above 65 will reach a total population of 1.5 billion in 2050, that will be about 16% of the world population, and that most of these individuals will come from developing countries. The older population in developing countries was projected to increase more than 250% in the period of 2010 to 2050, and in 71% in developed countries for the same period. In fact, the WHO report stated that by the middle of the current century, the elderly population above 80 years old in China could account for 100 million individuals, which is a considerably big population taking into account that last century there were just about 14 million that age in the complete world. The same tendency shows the next most populous country, that is India. According to this year's news on aging from the WHO, it is expected that by 2050, two-thirds of the world's population over 60 years will be living in low- and middle-income countries [2].

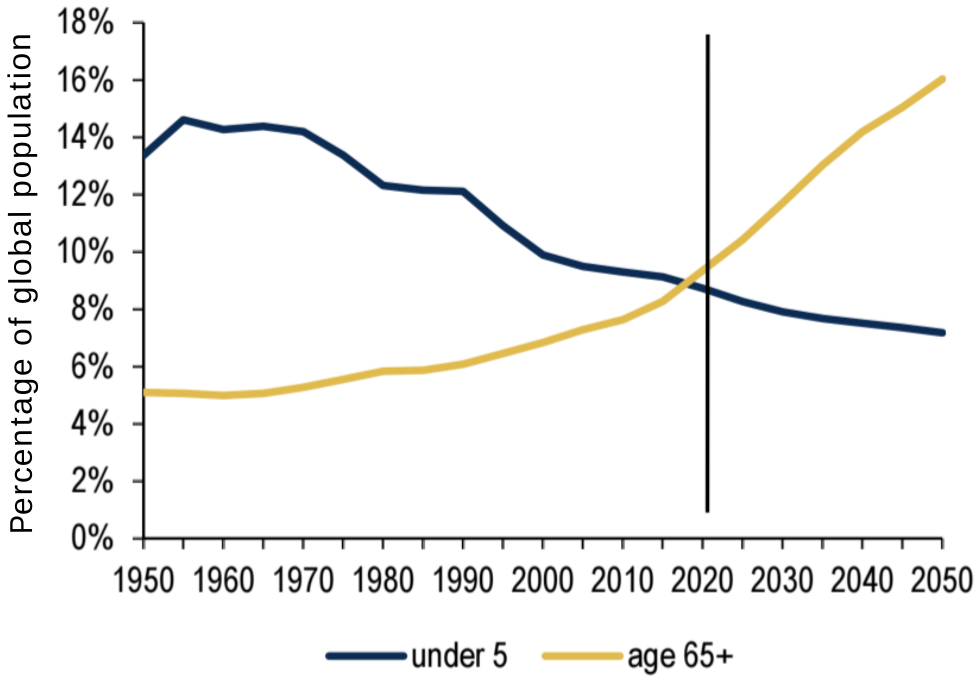


Fig.1.1 **Young children and older people as a percentage of global population.** Taken from the World Health Organization report on global health and aging, 2011 [1, 3].

Currently life expectancy in Japan, that is the actual leader, surpasses 83 years. Other countries as the case of Japan, except from some areas of Africa, have showed also an increment in life expectancy over the years, and for several it is at least 81 years. Experts estimate that this increment in life expectancy will reach an upper limit and stabilize or decrease; however, by 2010 it was still not clear if the life expectancy has reached this top point even in Japan, that along other countries like Sweden, have showed a continuous increment since reports from the year 1840, as shown in Figure 1.2 for the case of female population. An increment of 351% of the global population over age 85 was projected between 2010 and 2050, which is much higher than the projected increment of 188% for the population over age 65 and 22% for the population below age 65.

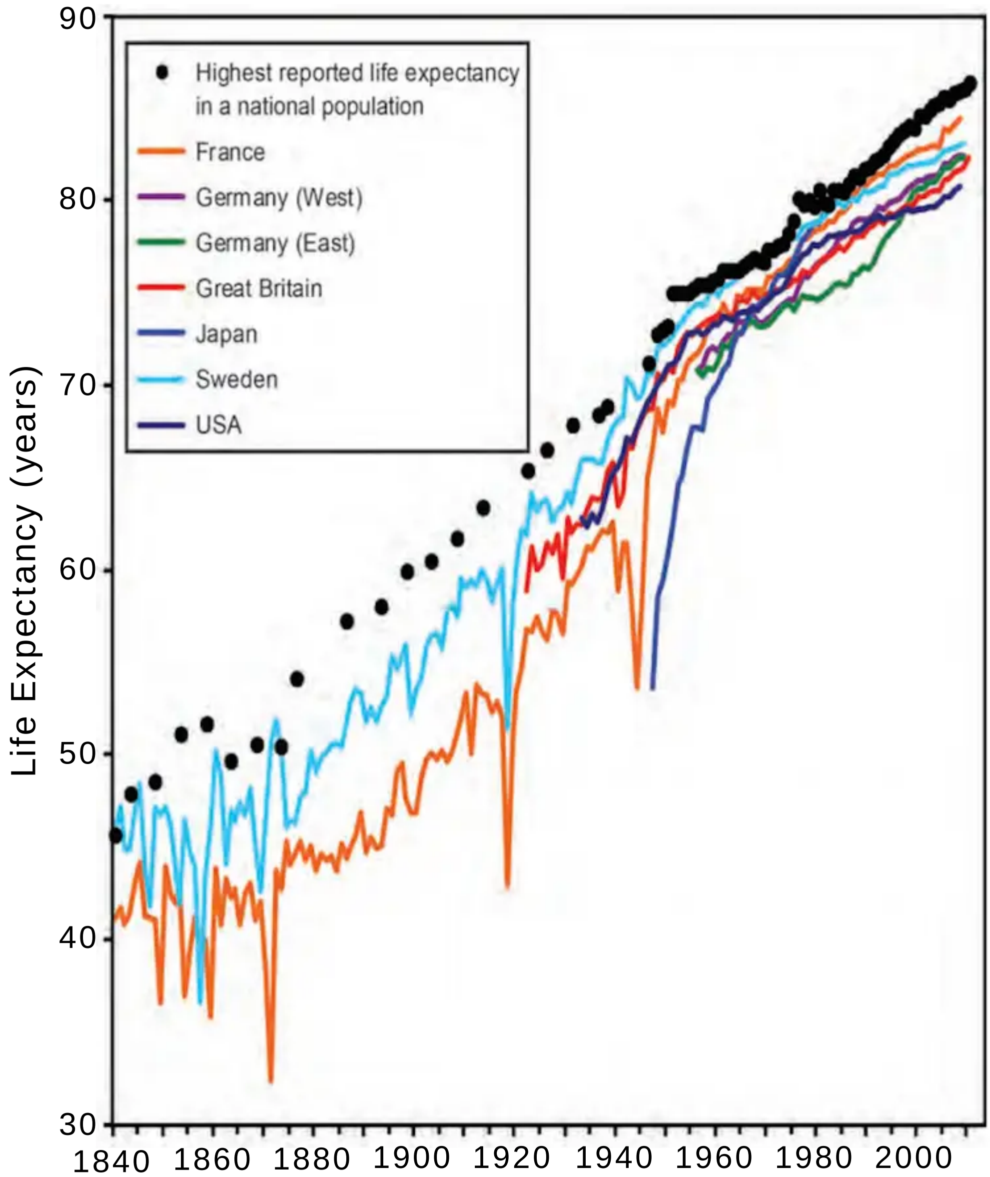


Fig.1.2 **Female life expectancy in developed countries.** Taken from the World Health Organization report on global health and aging, 2011 [1].

The current health reality of the older people show that instead of having a longer life with good health and well-being, their lives will be accompanied by illness and disability. In other words, if no right actions are taken to change this tendency, this old population will not rejoice in having a longer life of productivity and social interactions, but will have to withstand a longer life of dependency. This situation will be greatly detrimental for the health systems around the world, and evidently the costs to assist and care for the older individuals will represent significant investments from the health systems, that most probably several will not be able to afford and will collapse. The key for the health systems to be able to cope with the required assistance in the coming years is to focus from now on reducing severe disability from the health conditions of the older individuals because the longer these individuals can remain active and mobile, the more will they be able to care for themselves; this in turn will reduce the need of constant supervision by the healthcare providers and the need of specific facilities and life support machines.

1.1.2 Health afflictions in the elderly

The tendency to a longer life and the advancement of medicine have changed the principal health afflictions and causes of death of the older population. The world is currently in the time where the infectious and acute diseases have reduced, giving way to degenerative and chronic diseases to emerge. One of the most common degenerative diseases accounts for Alzheimer's disease, the most common type of dementia. According to the WHO [1], the probability to suffer from Alzheimer's disease doubles with every five years of age after age 65. It has been estimated that about 30% of elderly people above age 85 suffers from dementia.

The so called ”noncommunicable diseases” that are chronic diseases that include heart disease, cancer and diabetes are a major public health concern according to the WHO, since every region in the world presents every year higher incidents of death and disability caused by such diseases. One of the main factors that increase the risk of having a noncommunicable disease in old age is the lack of physical activity. An analysis from the National Institute on Aging (NIA) on the prevalence of chronic disease and disability among men and women aged 50 to 74 years, in the U.S., Europe and England, has shown that the portion of this population suffering from mobility impairment is very significant, as it can be observed in Figure 1.3. This means that such great amount of the older population are deprived from obtaining the sufficient physical activity they should in order to live in good health, and consequently the risk to have a chronic or degenerative disease is high.

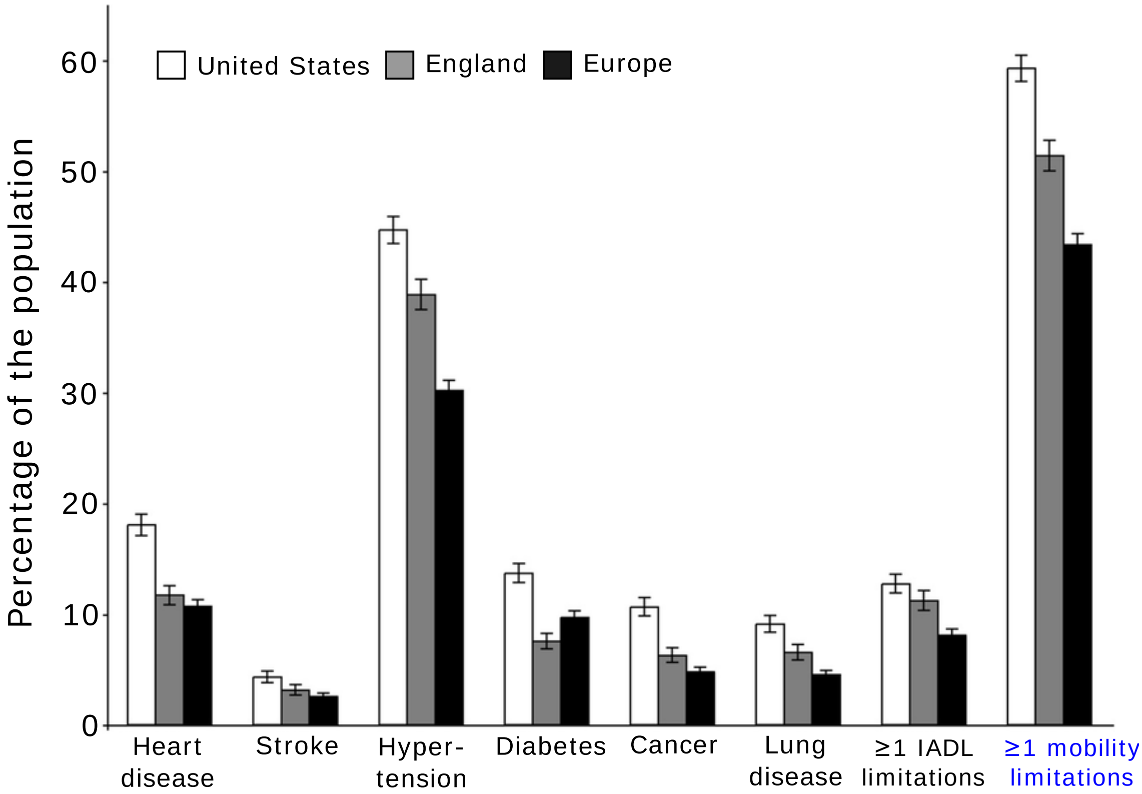


Fig.1.3 **Prevalence of chronic disease and disability.** Analysis among men and women aged 50 - 74 years in the United States, England and Europe in 2004. Taken from the World Health Organization report on global health and aging, 2011 [1, 4].

The loss of mobility affects directly the independent living of the older population and limits them to a life with a certain degree of dependency. To obtain more mobility, the older individuals are confined to the use of mobility aids like wheelchairs, canes and crutches. Such conditions make it for them harder to continue doing their normal activities of daily living (ADLs), including frequenting family and friends, which means having a poorer quality of life. In fact, ADLs, such as dressing, eating, bathing and walking become challenging. A recent report from the NIA stated that two-thirds of the older adults that leave hospitalization every year go through difficulties in their ADLs, including walking [5]. In addition, falls and disorders that affect their movement, such as Parkinson's disease are major causes of mobility loss.

1.1.3 Causes of mobility loss and prevention

The loss of mobility can be the result of a chronic disease, such as heart disease, pulmonary disease, diabetes, between others, and also the result of injury and sarcopenia (loss of muscle mass and strength with age). The older population who does not suffer from any chronic disease is still highly susceptible to the loss of mobility because of aging consequences that include loss in muscle strength and mass, changes in gait that affects their balance, as well as stiffer joints. In fact, sarcopenia is one of the main causes of functional decline in older individuals since with advancing age the body undergoes physiological and morphological changes in skeletal muscle that results in the reduction of the number and size of muscle fibers [6]. By age 80, up to 50% of muscle mass is already lost just by the result of this natural effect of aging [7]. With insufficient physical activity, sarcopenia in older adults might become even more severe, and result in a certain degree of mobility disability, which is defined as the inability to walk a quarter of a mile, or four blocks [5].

Moreover, injuries in older adults, that are very likely the result of falls, are a major public health concern because of the high risk they represent in impairing their mobility. In fact, injury is the fifth leading cause of death in the elderly, and most of injuries are caused by falls [8, 9]. Falls account for over 80% of injury-related hospitalization cases in the older population above age 65 [10, 11]. As mention before in this section, the NIA reported that most of the older adults that are dismissed from a hospital do not regain their total mobility, and even have trouble in accomplishing their normal ADLs. This means that after an injury a certain degree of mobility loss is highly probable.

It is clear that the prevention of injuries and slowing down the rate of sarcopenia are key actions to guarantee sufficient degree of mobility in the older age and diminish the mobility loss that can only lead to a poor quality of life and dependency. In addition, when older individuals already have a certain degree of mobility impairment, it is important that they receive the sufficient support from physical and social environments in order to regain the ability to perform their important activities, despite the impairment. Physical and social environments should encourage them to keep mobile by, for example, making the necessary adjustments to public buildings, transportation and places in general in order to allow a safe and easy ambulation and reinforce recovery. In this regard, the NIA-supported projects CAPABLE (Community Aging in Place, Advancing Better Living for Elders) and ABLE (Advancing Better Living for Elders) have provided to older adults home-based occupational and physical therapy, as well as home adjustments to ease the accomplishment of ADLs [5]. Such projects have showed to be effective for older adults to regain mobility and maintain independence, as well as to reduce considerably the economical and physical burden for their families and health systems. Currently, a total of 2500 low-income older adults in the United States and Australia have participated in these programs, and based on the successful results, such initiatives should be replicated allover the world.

It is well know that physical activity (regular exercise) is one of the main factors to live a healthy life. It improves people's physical performance, but can it prevent mobility disability in older age? The project LIFE (Lifestyle Interventions and Independence for Elders) asked this question, and assessed 1635 elderly people aged 70 to 89, who lived a sedentary lifestyle and were at high risk of major mobility disability [12]. The subjects were randomly assigned to a physical activity program or to a health education program. The physical activity program consisted in walking, strength and balance training, and flexibility. The health education program consisted in giving talks about nutrition, how to travel safely and body stretching and flexibility exercises. After 2.6 years of study, the results showed that the risk of incurring in a major mobility disability reduced in 18% for the case of the subjects that participated in the physical activity program, so the ones that did regular exercise, in comparison to the subjects that were involved in the health education program, who did receive some exercise but minor. This clearly shows the importance of keeping older adults in an active physical life, and to do so it is imperative that the conditions allow a safe ambulation with low risk to incur in a fall.

1.2 Falling in the elderly

Falls are detrimental for the elderly, for they are one of the major causes of pain, fractures, disability and even death in this population. About 30% of older people aged above 65 that live at home, and 50% of those living is healthcare facilities experience a fall every year, and 50% of those who have fall have recurrent falls [13, 14]. This means that the likelihood of incurring in a fall is high for older adults. After a fall, 20% of the cases require medical attention, and 5% of the falls imply a serious injury that include joint dislocations, severe head injuries and fractures [10, 15]. The probability that women above age 75 experience a fall is more that twice than that for men [16]. Additionally, 25% of the older population that are hospitalized for fall-related injuries die in the period of 12 months [17]. The reason of the death after a fall can be the severity of the injury, the mobility disability that result after the injury and hospitalization, and the mental and physical afflictions that such situation of disability causes.

The effects of experiencing a fall do not end with the physical injuries, but continue with the fear of falling again, which induces the older population to keep distance from social interactions and remain at home, and in severe cases they lose confidence and independence, and are even admitted to care institutions, where many end having severe depression [18]. In fact, the probability that elderly fallers enter a healthcare institution is twice the probability for non-fallers [19].

1.2.1 Causes

The risk factors that lead to a fall in the older population are related to their intrinsic characteristics and to extrinsic circumstances of the environment where they interact [20]. The intrinsic risk factors of the elderly that can lead to a fall include advanced age, poor health, female gender [21, 22], lower limb muscle weakness [23, 24] and functional impairment [25], visual impairment [26], changes in gait [27], problems in balance and posture [28], neurological impairment [29], musculoskeletal impairment [29, 30], sensory impairment [26], and the intake of certain medications [22, 31].

Lower limb muscle weakness, that most probably is the result of the loss of muscle mass (sarcopenia), increases the number of falls of older adults about two- to tree-fold [32, 33]. The loss of strength in the quadricep muscles has been found to be significantly related to the risk to fall [33]. The frequency of experiencing a fall due to lower extremity weakness is higher for institutionalized older adults than for those living in the community [34]. Moreover, the loss of somatosensory information at the hip and ankle can greatly increase the fall risk [35]. Visual impairment, as well, is related to the increased risk of falling, especially when the contrast sensitivity and depth perception are impaired [36]. To add, it has been also suggested that the lack of vitamin D is related to the fall risk of older adults because this vitamin is directly related to bone density and muscle strength [37].

Studies on falls in the elderly have reported that most occur during walking [38, 39], and that the principal causes of the falls are tripping and slipping [40, 41] in indoors [42], and outdoors activities [43]. A study that examined for a period of three years 130 older individuals residing in a nursing home [42] reported that the major cause of the recorded falls corresponded to incorrect weight shifting, that corresponded to 48% of the falls. The following causes were tripping and slipping that accounted for 21% and 3%, respectively. Additionally, this study reported that when taking into consideration the cause of fall and the activity doing at the time of the fall, tripping or stumbling while walking forward gathers the major of falls that corresponded to 11% of falls. Regarding the major causes when walking outdoors, Kojima et al. [43] screened 849 older subjects of both genders, and reported that slipping was the major cause of the falls corresponding to 48% of the falls, followed by tripping with 31.8%, and misstep with 4%. This study also reported that for older people above age 75 the majority of the falls were due to intrinsic factors.

A slip occurs when the friction force between the foot and the walking surface does not provide enough traction to counteract the shear force that is being applied by the foot. The applied shear force to the walking surface is at maximum at swing-stance or stance-swing transition, and it is at these times that slips are most likely to occur [44]. The higher the foot horizontal (anterior-posterior) velocity at the time of heel contact is, the higher the shear force and therefore the higher the probability to slip [45].

Trips occur when the foot is not able to clear the walking surface or obstacle during swing phase. The failure to exert a motion compensation to balance the body after tripping will potentially result in a fall. During mid-late swing, when the toes are closest to the floor (minimum toe clearance), the probability to experience a trip is the highest [46]. During walking, the minimum toe clearance, also known with its abbreviation as MTC, corresponds to the minimum value of the vertical distance between the walking surface and the lowest point of the toes of the swinging leg [47]. The MTC of healthy young adults corresponds to a value within 10 to 20 mm, and at this instant, the foot travels at its maximum horizontal velocity that is about three times the walking speed [47, 48]. Therefore, not being able to clear the ground at the time of the MTC most likely results in a fall [49]. To add, older adults present a lower MTC with higher variability when compared to young adults [49]; mainly due to ankle deviation known as drop-foot. A MTC of 12.9 mm has been reported in a study on a group aged about 70 [50], and a MTC value of 7.1 mm has been reported in a study on a group aged about 72 [46]. Hence, the MTC of older adults might exist within this range.

Moreover, instability while standing and walking is also a risk factor for falls in the older populations [51, 52]. Such instability has been found to be due to a medial-lateral imbalance that comes from the medial-lateral placement of the foot [53, 54]. For this reason, older individuals who are found to have a higher step width are prone to fall [55].

1.2.2 Prevention

The knowledge on the major cases of falling in the elderly has helped to device tools that are useful to determine if an older individual is prone to fall [56, 57, 58, 59, 60, 61], and has served as a baseline in the implementation of intervention programs that aim the prevention of falls [62, 63, 64]. In fact, the well-known Tinetti-test [56], or also called Performance-Oriented Mobility Assessment (POMA), is used to assess in older adults balance and stability during ADLs, and their risk and fear to fall. This test offers better prediction in regard to fall risk than other, also well-known, tests like Timed Up and Go test, functional reach test, and one-leg stand test [65].

In regard to intervention programs, a review [63] on numerous initiatives that promoted physical activity in older adults has shown that exercise programs oriented to the needs of every individual do evidently reduce the risk to fall. Even untargeted exercise programs have shown to be beneficial for fall prevention, especially programs that include Tai Chi or other exercises that stimulates good posture and balance; however, less effective than targeted programs. Thus, it is important to understand the main fall risk of every individual in order to emphasize the exercises that can help reduce those specific risks. The older population should be regularly screened to identify relevant intrinsic and extrinsic risks. It is recommended to not engage elderly fallers in exercise programs that include walking; instead, balancing and strengthening exercises should be the focus [66]. Additionally, not only the older adults at high risk to fall should be involved in such physical intervention programs, but the older community in general, because all will profit improved strength and functional capabilities [66]. The guidance and motivation for a trainer has been found to be important because about 50% of the older adults that are offered an exercise program refuse to participate [67].

Moreover, conducting a hazard assessment at home and making the respective adjustments to minimize those hazards have proven to be effective to prevent falls [68, 69]. The effectiveness appear to be higher when the home adjustments are delivered by an occupational therapist [68]. The older population that mainly profit from such assessment and modifications at home are the ones with already a fall history and mobility impairments [70]. This intervention at home should not be the only one implemented with the objective of diminishing the risk to fall of an older adult, but should be combined with other initiatives like physical intervention programs that can promote recovery, reduce the fear to fall, and help to maintain an independent life at advanced age.

Multicomponent intervention programs, where the older adults participate in two or more interventions that are fixed (not adjusted to the specific need of every individual), have also shown to significantly reduce the number of fallers and the frequency of the falls [71]. It has been controversially discussed if such programs designed for the specific needs of every older individual would be more efficient and deliver better results in the prevention of falls [72]. The individual design would represent much higher costs and the need of more operational practitioners. Additionally, The reduction of psychotropic medications intake has shown to significantly reduce the fall risk in older adults, especially the slow withdrawal [68, 73].

When going much deeper on the analysis of intrinsic characteristics and finding out the source of the falls, it is the gait cycle of the older population where we have to look at. In the gait cycle, the critical factors in regard to fall prevention are situated in the swing phase. More specifically, toe off, that marks the beginning of the swing phase, and heel contact, that marks its end, are the key gait events for preventing slips, and the MTC, that occurs at mid-late swing, is the key gait event for trip prevention [49]. If trips and slips are prevented in older adults, the occurrence of falls will diminish considerably, as tripping and slipping account for the main fall initiators.

1.3 Healthy elderly gait: age-related impairments and compensations

Before going into detail on the gait cycle, let us have a short introduction on the different phases that are used in literature to describe it. The gait cycle is usually broken down into several phases and instant events, as shown in Figure 1.4. When looking the gait cycle from the point of view of one single leg, it is broken down into the stance phase, that corresponds to the time when the foot is contacting the ground, and the swing phase, that is the period when the foot is in the air swinging. When observing from a bilateral viewpoint, the gait cycle is broken down into the single support phase, that is the period when only of the feet is contacting the ground, and the double support phase, that is the period when both feet are contacting the ground. In regard to the instant events usually referred to when talking about the gait cycle, we can list the following: heel contact, that is the instant when the heel contacts the ground to initiate the stance phase, toe off, that is the instant when the toes leave the ground to initiate the swing phase, and MTC, that is the instant when the toe clearance reaches its minimum value in mid-late swing. Please refer to Figure 3.2 in Chapter 3 for other more specific sections that are also used to describe the gait cycle in more detail.

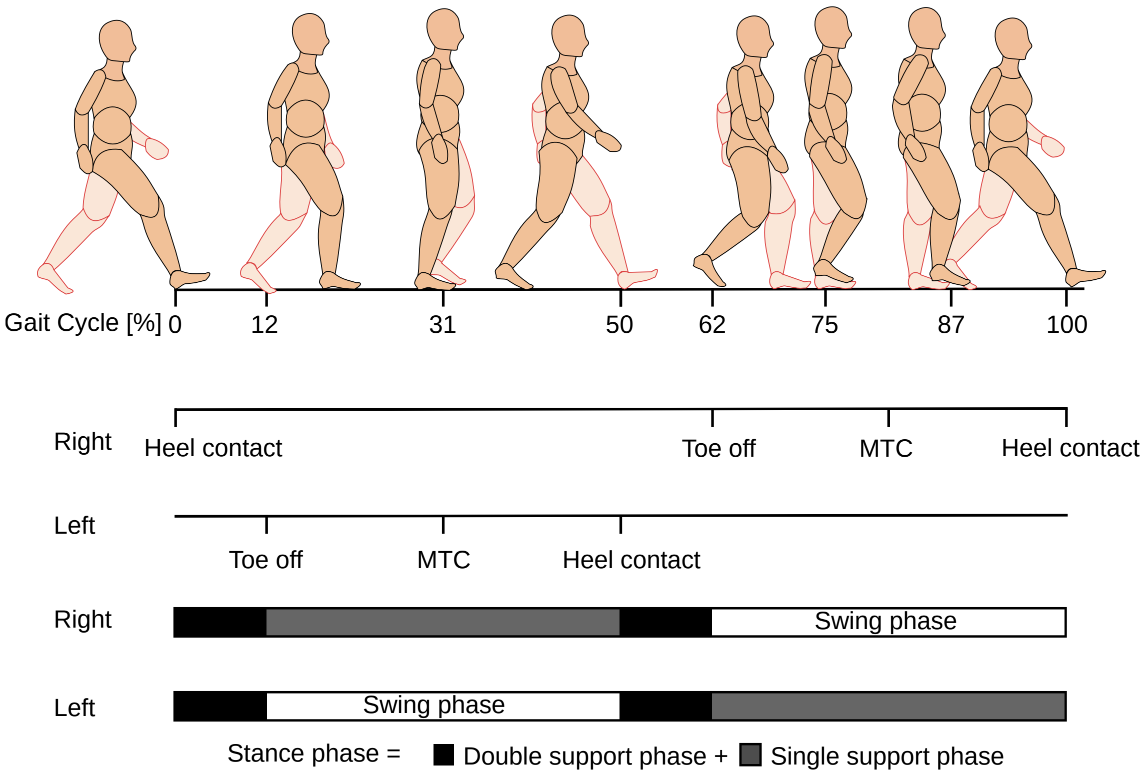


Fig.1.4 **Gait cycle.** Characteristic gait phases: stance phase, swing phase, single support phase and double support phase, and gait events: heel contact, toe off and minimum toe clearance (MTC).

1.3.1 Spatio-temporal gait parameters

The common spatial parameters that are usually used to characterize the gait cycle are the step length, step width, stride length, and the temporal parameters are the duration of the single and double support phases, of the stride, and of the stance and swing phases. Another used temporal parameter is cadence, that is the number of steps taken per minute.

Studies have shown that the preferred walking speed, that can be calculated dividing the stride length by the stride duration, of older adults is significantly lower than that of the young [74, 75, 76, 77]. The preferred walking speed reaches its peak at the age of

45, and from that point it starts to decrease [78]. The reason of the reduction in the self-selected gait velocity in the elderly is the decrement in the stride length [74, 79, 80]. To add, there is no consensus in regard to the stride duration, several studies have found in the elderly a longer stride duration in comparison to the young [75, 78, 81], others have reported no significant differences [76, 82, 83], while others have reported a shorter stride duration [79, 84].

The elderly show also a longer double support duration than that of the young, which is the result of spending less time in the swing phase [55, 74, 81]. Moreover, some older individuals show a higher step width, that causes higher medial-lateral instability [55]; however, it is not found in the majority of the elderly, and it is suggested not to be a effect of natural aging [75, 80, 85].

Several researchers have suggested that the reduction of the duration of the single support phase, and the consequent increment of the duration of the double support phase, and the reduction of the stride length are adaptations of the elderly to a more stable gait [55, 84, 86]. Other researchers have suggested that the lower preferred walking speed of the elderly and the shorter stride length are signs of aging and the loss of functional capability [87].

1.3.2 Changes in the lower limb kinetics and kinematics

Toe clearance

In the research community a clear focus has been placed on the minimum value of the toe clearance or MTC because of the relationship to the risk of tripping in the elderly as described in the previous section ”Falling in the elderly”. The MTC has been found to be significantly sensitive to hip adduction-abduction during the stance phase. Thus, 1 degree difference in the adduction-abduction angle results in a reduction of 5 mm in MTC [45]. It has been also reported that the MTC is twice as sensitive to the plantar-dorsal flexion of the swing ankle in comparison to the flexion-extension angle of the swing hip, and that the MTC is considerable insensitive to the flexion-extension angle of the swing knee [88]. A difference of 10 degrees of knee flexion-extension angle is necessary to variate the MTC in 5 mm [88].

The MTC in young adults has been found to be about 10 to 20 mm citeBarret10, Khandoker10, and researchers agree in saying that the MTC of the young is approximately 15 mm [46, 74, 75, 88]. In the case of the elderly, an MTC of 7.1 mm was reported by Begg et al. [46], and a value of 12.9 mm was reported by Karst et al. [50]. Controversially other researchers have reported no significant difference on the MTC between the young and the elderly [74, 80, 89]. Since the control of MTC is highly redundant, for the MTC depends on a seven segment kinematic chain that comprises the pelvis and bilateral foot, shank and thigh segments, several kinematic compensatory strategies might be applied by the elderly [49]. Thus, ought to such strategies, the elderly could be able to maintain the MTC, and this is a possible explanation of why several studies did not find any difference on the elderly MTC when compared to that of the young.

Heel contact velocity

The heal contact velocity is another change found in the elderly gait, and is suggested to a major cause of slipping in the elderly, as described in the previous section ”Falling in the elderly”. The anterior-posterior velocity at heel contact has been found to reach zero velocity at about 50 ms after the contact in the case of not slipping [90]. A velocity higher than that would indicate a slip.

A higher anterior-posterior heel contact velocity has been reported Winter [86] for the elderly even when in this study the elderly walked at a preferred speed lower than that of the young. In this study, it has suggested that the reason behind the higher heel contact velocity is a trend in the elderly to a lower knee flexor power absorption at the terminal swing phase. The study of Mills [49] agreed with the higher anterior-posterior heel contact velocity in the elderly, and suggested that the reason of such increment in velocity is a lower swing shank angular velocity at the time of heel contact.

Ankle joint

Regarding the ankle joint, there is a consensus in the findings about the age-related changes in the kinetics and kinematics. Various studies have found a lower ankle joint ROM in the elderly while walking at preferred speed [74, 75, 79, 81, 84, 91], also when walking at a similar speed to that of the young [92], and at a faster speed [91]. This reduced ROM is suggested in these studies to be due to an age-related lower ankle plantar flexion peak.

Moreover, a reduced ankle plantar flexion power generation at terminal stance has been found in the elderly [74, 79, 91, 93, 94], that is the result of an age-related lower plantar flexion moment [86, 92] and/or a lower plantar flexion rate [81, 86]. The plantar flexion power at late stance is regarded as the power that most contributes to the forward propulsion of the body [95]. To add, it is suggested that this reduction of positive work at the ankle in the elderly is related to the found increased energy given by the hip joint during early stance and the transition between the stance to swing phase.

Knee joint

At self-selected walking speed, an age-related reduction knee flexion peak in the swing phase [75, 80, 96], a consequent lower knee flexion-extension ROM [79, 80], and a higher knee flexion at the time of heel contact [75, 86] have been found in the elderly gait. The more flexed knee at heel contact might result in a lower step length [86]. In agreement to this findings, Kaneko et al. [81] found in older individuals aged about 80 years a higher flexed knee at heel contact and toe off at a fast walking speed in comparison to individuals aged about 50 years. It has been also reported a lower knee flexion peak during swing phase [75].

Some studies have reported no differences between the elderly and the young in the knee joint power during early middle stance [74, 79], while others have found an age-related reduction when walking at a preferred speed [91] or at a set speed [92]. Similarly, some studies have reported no difference in the knee power at terminal stance phase in comparison with the young [79, 97], in another study a lower knee extensor power absorption was found in the elderly [91], while another reported a higher power absorption [86]. The knee power at early middle stance and terminal stance is important to slow down the rate of knee flexion that would otherwise abruptly occur due to gravitational and inertial moments.

Winter [74] found in the elderly a trend to a lower knee flexor power absorption at terminal swing; however, this difference to the young was not statistical significant. The knee power absorption at terminal swing is known to contribute to a safe heel contact since it acts to lower the foot velocity before the contact. Winter suggested that the trend to a lower power absorption is the reason why the elderly have a higher risk to slip due to a higher foot velocity at the time of heel contact.

Hip joint

There is no consensus in regard to the hip flexion-extension range of motion (ROM) for the elderly. Several studies have reported a lower ROM [75, 81],others have found not significant differences with that of the young [80, 96], while other have found a greater ROM [79, 86]. Furthermore, a lower hip hyperextension peak has been found in the elderly [91, 98, 99], that is suggested to be the result of a higher anterior pelvic tilt [91, 99]. The higher pelvic tilt has been reported to be an age-related tightness or contracture of the hip flexors [99]. Finley et al. [84] reported a greater hip abduction-adduction ROM for the elderly , and in agreement Judge el al. [79] reported a greater hip adduction for the elderly during middle to terminal stance.

A higher positive hip extension work during loading response and a lower negative hip flexion work during middle stance have been reported for the elderly [97]. The reason of the increment in the hip extension power at the stance phase is suggested to be a compensatory strategy of the elderly for an age-related decrease in the generation of ankle plantar flexion power at terminal stance [92]. A lower hip adduction power peak in the stance phase and a lower hip adduction work at terminal stance have been suggested for the elderly in comparison to that of the young; however, the differences were found not to be significant [97].

1.4 Understanding the elderly gait for the development of gait assistance devices

As evidence given before in this chapter suggests, promoting physical activity in the elderly is of main importance to preserve independence, and thus a good quality of life. One approach to achieve this is by the means of assistance devices. A major challenge is the planning, designing and testing of such devices because all this procedure represents high injury risks for the older adults, and due to health and ethical considerations, these individuals are limited to participate. Thus, methods are needed that define how, until what extend the participation of the older individuals can be replaced from such procedures, and who can take their place. The project SiNuS-Pflege (Simulation of Nudging methods to Strengthen the independence of people in need of care) is currently investigating how far assistance approaches can be tested without involving people with health impairments. A method to carry out such tests without the participation of the elderly is through age simulation suits or also known as silver simulators, whose main objective is to simulate elderly impairments in the young. In mid-2000, the project SiNuS-Pflege carried out a workshop to determine until what extend the commercialized age simulation suit GERT (GERonTologic test suit) can be used to gather reliable data that can represent the elderly. The work on this area of age simulation suits has still a long way to go, especially when talking about the simulation of certain impairments of the elderly in real values.

1.4.1 Age simulation suits

Several age simulation suits have been devised till date, and they include: the ”Third Age Suit” from the Ford Motor Company, the ”AGNES” suit from MIT [100], the ”GERT” suit [101] and the ”R70i” exoskeleton [102]. The main objective of these suits is to sensitize the young, primarily medical students and engineers, on elderly impairments that include: restrained joint ROM, limited dexterity, and visual and tactile sensory impairments, so that these professionals can provide better services and care for the elderly.

It has been suggested that in order to simulate natural sensations of aging, age simulation suits should be composed of: (a) a head module, to reduce vision and hearing capabilities and neck mobility, (b) an arm module, to reduce tactile capability and shoulder, elbow and wrist mobility, (c) a torso module, to reduce spine mobility, and (d) leg modules, to reduce the hip, knee and ankle joints mobility [103].

GERT suit

The GERT (GERonTologic) age simulation suit offers to the young the opportunity to experience age-related impairments that comprises narrow visual field, opaque sight, loss of high-frequency hearing, restricted head mobility, stiff joints, loss of strength, lower grip ability and lower coordination skills [101]. This suit is a commercial product and it has been made in modules to ease the combination of the required ones to simulate specific impairments according to the application. People with different body sizes can use GERT. As shown in Figure 1.5a GERT suit contains a neck brace, a weight vest, several bandages for the elbows and knees, weight cuffs for the ankle joints and wrists, a pair of gloves, goggles, and ear plugs [104].

AGNES Suit

Figure 1.5b shows the AGNES (Age Gain Now Empathy System) suit, which was devised by the Age Lab at MIT. It uses straps, braces and bands that can be customize to meet the body size of a specific individual. AGNES can reduce the joints ROM (wrist, elbow, shoulder, cervical spine, knee and hip), balance and visual, hearing and tactile ability, and can also simulate muscle fatigue. According to AGNES developers, it has been calibrated to simulate equivalent impairments of older adults in their mid-70s. The suit is equipped with modified shoes to impair balance, a variety of gloves and goggles to simulate different degrees of sensory impairment. When testing the suit in young adults, it was validated that they experienced changes in task performance consistent to age-related impairments of the elderly [100]. Feedback from the wearers have emphasized the better understanding of how it is to walk in the shoes of an older adult, and the better empathy they have towards this community.

Fig.1.5 **(a)** GERT suit [101, 104], **(b)** AGNES suit [100, 105].

Third age suit

Figure 1.6a shows the Third Age Suit, which is also called the ”empathy suit” from the Ford Motor Company. Ford has been implementing the suit since 1995 [106]. It was devised for engineers and car developers to help them realize how to be old is, and to get conscious about the limitations of the older community, so that at the time of engineering a vehicle, they take into account the needs and limitations of this community. The suit is made of nylon and is equipped with elbow and knee braces to simulate arthritis, a neck brace to limit it mobility, active gloves to reduce finger strength and simulate tremors (like the ones someone with Parkinson would experience), weights on the feet to limit lifting them, a pair of ear muffs and headphones to simulate hearing issues, and a pair of glasses to impair the sight and simulate glaucoma and cataracts.

R70i exoskeleton

Figure 1.6b shows the R70i exoskeleton [102], which has been produced by Applied Minds LLC for Genworth. It has active actuated joints to restrict their mobility. The resistance to the joints mobility can be adjusted, and the manufactures claim that young people can experience the effects of arthritis, the loss of muscle strength and also the feeling after having a hip replacement. The R70i also includes a pair of headphones that includes a microphone and a microprocessor to manipulate what the person hears. In that manner, it is claimed that the R70i can simulate in the young the loss of hearing and tinnitus (ringing of the ears). The suit can also simulate the difficulty to speak and form words (aphasia). In addition, the R70i is equipped with augmented reality goggles with cameras on the front that captures live what the person is looking at. The video is manipulated in real time, and in such manner the wearer can live the effects of impairments like glaucoma, macular degeneration and cataracts.

(a) (b)

Fig.1.6 **(a)** Third Age Suit [107], **(b)** R70i exoskeleton [102, 105].

1.5 Problems statements

With the rapid increment of older population in the world, it is a major public health priority to provide the needed assistance in order to guarantee a good quality of life and maintain independence. As described in this chapter, there are various age-related changes in the elderly gait that significantly increase their injury risk as consequence of falling. Wearable devices could do the job to overcome such gait changes and give back to the elderly the ability to walk safe and without fear. The planning, development and testing of such devices face vast of restrictions because of health and ethical concerns that arise due to the dangers each of these processes imply when involving elderly people in the needed experimental trials.

Methods to simulate specific gait impairments of the elderly in the young would make it possible to carry out until certain extend the mentioned processes, for engineering assistance devices, with the young instead of the elderly. Such methods could be the existing aging suits, that are also described in this chapter. However, these aging suits do not reproduce the elderly impairments in an accurate manner, and validation of using these suits to reproduce in the young data equivalent to that of the elderly is missing.

Before starting this doctoral project, we proposed MARTT (Muscle Activity Restriction Taping Technique) as a method to reproduce the MTC of the elderly in the young [108]; however, the technique was not validated in terms of reproducing real MTC values found in the elderly while maintaining the natural joint motion patterns (i.e. not causing strange patterns that do not agree with the characteristics of human walking).

In addition, compensatory motions are natural in an impaired gait. As described in this chapter, several age-related joint compensation motions have been suggested in studies on the elderly walking; however, there exists little knowledge about which of the gait changes due to aging correspond to compensation motions and which to impairments, and also how the compensatory kinematic chain works (i.e. what they compensate for and when).

1.6 Project aims and Hypotheses

The general aim of this doctoral project was to assess the feasibility of reproducing with MARTT in young adults age-related gait changes of the elderly . The following are the specific aims and hypotheses of this project:

Aim 1: Compare the MTC central tendency, variability and distribution of young male adults, whose lower limbs have been restricted with MARTT, with the respective values that have been reported in literature about the elderly.

Aim 2: Validate MARTT as a technique that is able to reduce the MTC of the young to the values of the elderly without causing strange walking patterns in the young that do not correspond to the natural human gait cycle.

Aim 3: Compare the spatio-temporal gait parameters of the young adults walking with MARTT with the age-related changes in these parameters reported for the elderly.

Aim 4: Determine the similarities between the gait compensation motions observed in the young adults walking with MARTT, and the age-related compensations reported for the elderly in literature that are a result of muscle weakness.

It was hypothesized that by restricting the transversal area of a muscle when it contracts, it would be possible to reproduce in the young the effects of age-related muscle weakness; more specifically, the reduced MTC of the elderly. In addition, since the method of blood occlusion has been used by athletes for training, it was hypothesized that MARTT will not impair the natural human gait cycle of young adults.

In addition, it was hypothesized that the joint compensation strategies caused by MARTT in the young would have similarities with the compensation motions that have been reported in literature as the result of age-related muscle weakness.

Chapter 2

Validation of MARTT technique to reproduce the MTC impairment of the elderly in the young

2.1 MARTT technique

Muscle Activity Restriction Taping Technique (MARTT) is a technique first introduced in the study of Ullauri et al. [108] as a method to reduce the minimum toe clearance (MTC) of the young, with the objective of reproducing in them the MTC values of older adults, while maintaining the natural motion patterns of the lower-limb joints. Figure 2.1 shows how MARTT was implemented in every subject and the muscles at the lower limb that it targeted to restrict. As depicted in the figure, MARTT implements several belts to exert a compression force at different areas of the lower limb. During muscle contraction, the applied compression force limits the transversal area of the muscles' belly, that in result bounds the changes in muscle length. In this manner, MARTT targets to restrain the joints range of motion (ROM) of the lower-limb to reduce the MTC.

Figure 2.2 shows MARTT belts, that are made of non-stretchable fabric to maintain a constant and a uniform muscle restriction at the area where they are applied. In order to reduce the ankle flexion and subsequently the MTC, a pair of MARTT belts were

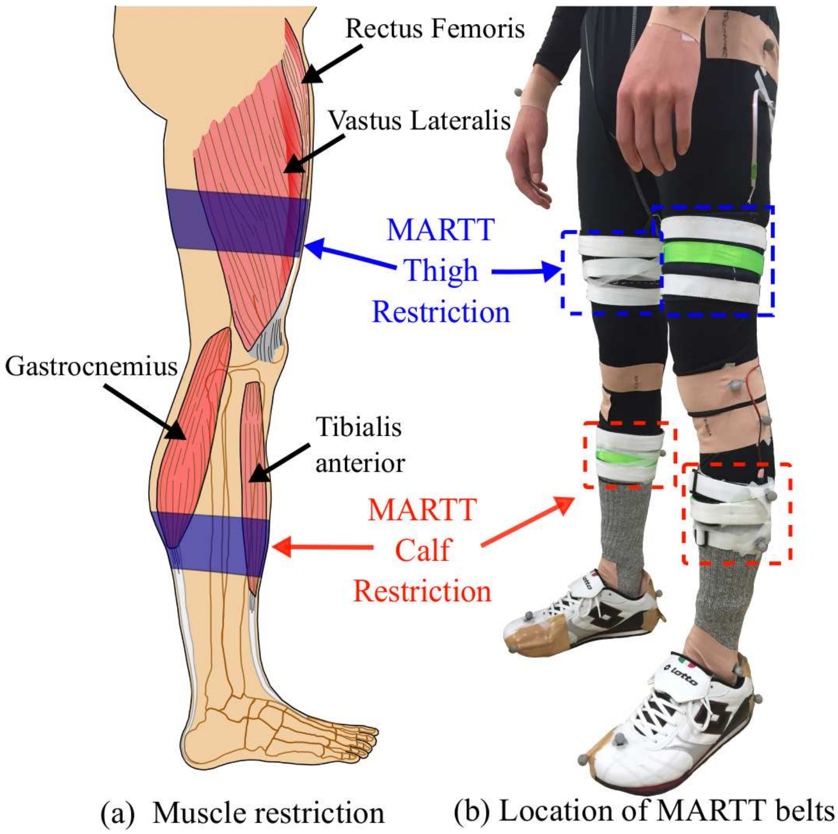


Fig.2.1 **MARTT apparatus.** a) Muscles at the shank and thigh that MARTT targets to restrict and (b) Location of MARTT belts at the lower limb.

applied at the shank area that restrict the tibialis anterior and gastrocnemius muscles. Additionally, in order to avoid compensation motions by the hip and knee joints, and emphasize the reduction of the MTC with a lower flexion at the hip and knee joints, a second pair of MARTT belts were applied at the thigh area to restrict the vastus lateralis and rectus femoris muscles.

The reference point to place the belts at the calf area is the end of the gastrocnemius muscle heads. This point is aligned with the lower edge of the belts. The reference point to place the belts at the thigh area is the beginning of the rectus femoris muscle. This point is aligned with the upper edge of the respective belts.

The above mentioned compression force implies blood flow reduction (also known as ”blood flow occlusion”) in the restricted muscles. In moderation, blood flow occlusion does not negatively affect the human body, in regard to endothelial function and blood

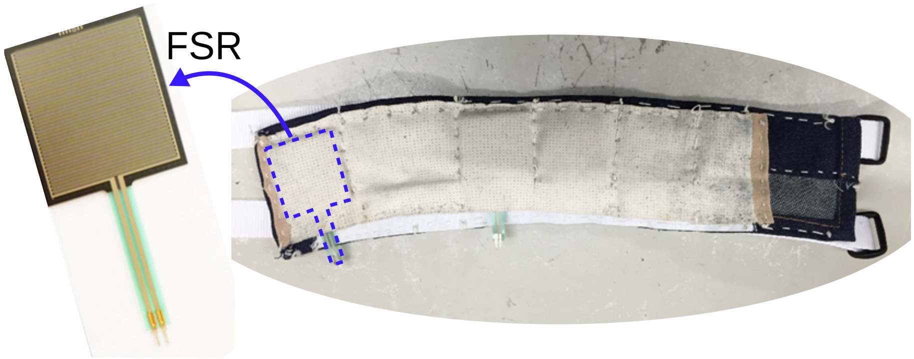


Fig.2.2 **MARTT belts.** Belts made of non-stretchable fabric, that integrate flexible force-sensing resistors (FSR) to monitor the applied force. A total of 3 FSR sensors were embedded in each belt.

coagulation [109]. Because of this no negative impact on the body, blood flow occlusion is even used for improving endurance [110]. As shown in Figure 2.2, MARTT belts incorporate flexible force-sensing resistors (FSR) to monitor and measure the applied restriction force on both legs, in order to ensure that a safe force value is being applied and that both legs are equally restricted. In this manner an asymmetric walking is prevented and the natural gait patterns of the young can be maintained. A total of 3 FSR sensors were embedded in each belt. The restriction force applied by MARTT belts was set to match a pressure of about 180 mmHg. A pressure value below 200 mmHg was selected due to the fact that a pressure range of 160-230 mmHg has been reported as a safe range for restricting the blood flow [111, 112], and a study that restricted leg blood flow in older individuals reported significant muscle fatigue at the pressure of 200 mmHg [109].

2.2 Hypothesis

In this study, it was hypothesized that by restricting the transversal area of a muscle when it contracts, by implementing MARTT, it would be possible to reproduce in the young the effects of age-related muscle weakness; more specifically, the reduced MTC of the elderly. In addition, since the method of blood occlusion has been used by athletes for training, and have been found not to be negative for the human body, it was hypothesized that MARTT will not impair the natural human gait cycle of young adults.

Thus, this study aimed at validating MARTT as a technique that can reproduce in the young the MTC values found in the elderly. For this purpose, the central tendency, variability and distribution of the MTC of young subjects walking with MARTT are the main focus of this study, as well as, verifying that MARTT does not cause strange gait patterns that do not correspond to the natural gait joint patterns of human walking.

2.3 Experimental setup

Subjects

This study evaluated ten male young-adult subjects with an average age of 22 years, and an average weight and height of 63.9 $\pm$ 8.9 kg and 1.72 $\pm$ 0.05 m, respectively. The right leg was the dominant leg of all subjects. All participants were carefully selected to guarantee that none suffer from any neurological or musculoskeletal impairment. Every subject was asked to promptly express the case of feeling pain or any other uncomfortable feeling that could lead to change their natural lower-limb joints' motion patterns.

The subject recruiting was carried out within the student population of Nagoya University, in October 2015. Before the study, each selected subject was informed, in a verbal and written manner, about the objective of the study and how it will be conducted, and each of them signed consent form.

This study was approved by the Institutional Review Board of Nagoya University and it was registered under the approval Number, 14-4.

Equipment

A motion capture system (MAC 3D system, Motion Analysis Corporation, U.S.) was implemented to record the gait cycle of every subject at a frequency of 100 Hz. According to the motion module the software for interactive musculoskeletal modeling (SIMM, Musculographics Inc., U.S.), reflective markers were positioned along the subject's body. This software was utilized for computing joint angles.

Figure 2.3 shows that to record with the motion capture system the toe clearances during the complete gait cycle, a reflective marker was placed right next to each hallux, centered at the tip of the distal phalanx. Similarly, a marker was attached to the back of each calcaneus to capture the heel clearances. In addition, as shown in Figure 2.4, flexible force-sensing resistors (FSR-400, Interlink Electronics, U.S.) were embedded in a pair of comfortable shoes to detect toe off and heel contact events of the gait cycle. In total two of these force sensors were embedded in each shoe, one located at the tip and the other at the sole of the shoe. Thanks to the characteristics of these sensors, they were not perceived by the subjects while walking and did not affect to the natural walking patterns.

Fig.2.3 **Shoe markers.** Markers used to determine toe (Mt) and heel (Mh) distances to the ground in every step.

The subjects walked on a treadmill with a 1.4×0.5 m^2 walking surface (OMEGA3, Johnson Health Tech Co., Taiwan). In each walking trial, the subjects heard a non-rhythmic noise from a pair of headphones in order to prevent any distraction from the noise generated at toe or heel contacts to the walking surface.

Protocol

The natural gait, referred from hereon as normal walking, of the subjects, as well as, their gait when being restricted by MARTT were recorded. Two types of restrictions were applied by MARTT and two walking speeds were studied. This makes a total of six

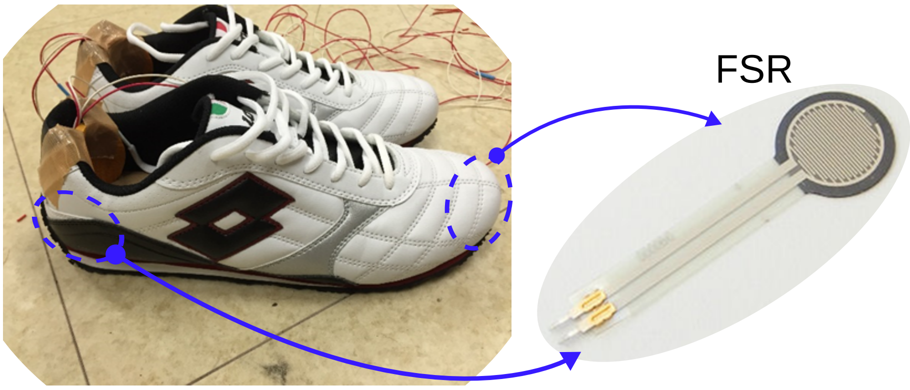


Fig.2.4 **Embedded shoe sensors.** Flexible force-sensing resistors (FSR), one located at every tip and sole of a pair of shoes (2 FSR in each shoe), were utilized to detect the toe off and heel contact events on every gait cycle.

walking trials carried out in every subject (3 walking conditions * 2 walking speeds).

The two walking speeds corresponded to a speed of 1.11 m/s (4 km/h) and a speed of 0.97 m/s (3.5 km/h). The speed of 4 km/h is suggested as an average of the natural walking speed in the young [46], whereas the speed of 3.5 km/h is suggested as an average speed for healthy older adults [113]. The two restriction conditions comprises the condition where MARTT belts were applied at the shank only, and the conditions where the belts were applied at both the shank and the thigh. The restriction of the shank muscles is referred from hereon as C-restriction, and the restriction of both the shank and thigh muscles is referred from hereon as CT-restriction.

The normal walking of every participant was recorded at the beginning of the study, and the walking speed was selected in a random manner. After this, a restriction condition (C- or CT-restriction) was applied with MARTT belts and the subject walked on the treadmill at the selected walking speed (3.5 km/h or 4 km/h). The combination of the restriction condition and walking speed was selected in a random order for each subject. During the complete duration of the study, when MARTT was utilized, the restriction forces being applied by MARTT bests were constantly monitored to make sure both legs are equally restricted and the restriction force is kept invariant. When changing from one restriction condition to the other, the applied restriction force was checked to always match the 180 mmHg pressure.

Data processing

The first minute of all gait recordings was excluded from the processing to make sure the adaptation time of each subject to the experimental condition (walking speed and MARTT restriction) has passed. The recorded motion data were filtered with a 6-Hz Butterworth filter and the position of the markers in the recorded motion data were linked to the respective locations in the human model of SIMM with which the joint angles were computed.

The data were cut in every gait cycle that was determined as the walking period comprehended between two consecutive heel contact events of the same leg. The heel contact and toe off events were detected by monitoring the instants when the normal force, measured by the force sensors located at the soles and tips of the shoes, increased. All data, including toe and heel clearances and joint angles, were normalized for each gait cycle. In addition, the toe contacts to the walking surface (zero toe clearance) during the swing phase were manually quantified while the subjects walked.

The orthogonal distance from the treadmill surface to the markers placed at the shoes right close to the hallux and calcaneus correspond to the toe and heel clearances, respectively. The MTC was computed as the minimum value of the calculated toe clearance during the middle of the swing phase of the gait cycle.

The median and interquartile range (IQR) of the MTC were calculated for each subject to give a measure of the central tendency and variability of the MTC since the distribution of the MTC was found to be leptokurtic and positively skewed, as also reported in literture [46].

The Mann-Whitney-Wilcoxon test was utilized to determine the significance of the reduction of the MTC in every subject. In order to analyze the reduction of the MTC between subjects, the MTC calculated for every subject was normalized to his height, and from these normalized values the mean, SD, median and IQR of the all-subjects MTC were calculated. The significance of the differences of the calculated between-subjects MTC during normal and restricted walking were statistically analyzed utilizing the one-tailed paired t-test.

For every subject, spatio-temporal parameters (cadence, single-support phase and step length) of the gait cycle and joint angles were analyzed with the one-tailed paired *t*-test to determine the effect of MARTT on these parameters in comparison to normal walking. The median and IQR values of spatio-temporal parameters between subjects were calculated to determine the their general tendency, and the significance of the found differences was analized by the Mann-Whitney-Wilcoxon test.

During the swing phase, the average toe clearance of every subject was normalized to their height, and the average toe clearance between subject was computed in order to observe the effect of MARTT on the toe clearance in comparison to normal walking.

2.4 Results on the reproduction of the MTC of the elderly

Gait timing

It was confirmed that after the application of MARTT in every subject, their symmetrical walking was maintained. That is, the gait cycle percentage between the heel contact event of one leg and the heel contact event of the other leg corresponded to 50 % on average for both walking speeds, as it is in normal walking. In regard to the spatio-temporal parameters, Table 2.1 shows the median and IQR values corresponding to the cadence, single-support phase and step length between subjects. This last was normalized to the subject's height before computing the between-subjects step-length value.

In Table 2.1, it is observed that the step length and cadence reduces with the walking speed. The statistical analysis revealed no significant differences between normal walking and restricted walking in regard to the cadence and step length, and a significant difference in regard to the single-support phase. During C-restriction, for both walking speeds, eight subjects exhibited a single-support phase significantly ($p < 0.05$) shorter than that in normal walking. Similarly, during CT-restriction eight subjects at the walking speed of 3.5 km/h and nine at the speed of 4 km/h exhibited a significantly ($p < 0.05$) shorter single-support phase.

Joint motion

With the objective of analyzing the effect of MARTT in the lower-limb joints, the ankle, knee and hip joints ROM were calculated for every subject, as well as, the maximum plantar flexion of the ankle joint during the gait cycle and the knee flexion angle at the end of the swing phase. The joints ROM were calculated in percentage of the ROM during normal walking for every subject.

Figure 2.5 shows the distribution of the between-subject joints ROM, maximum plantar flexion and knee flexion, previously mentioned, for both walking speeds.

Table 2.1 **Gait timing between subjects**

Parameter	Walking Speed [km/h]	Normal walking	C-restriction	CT-restriction
Cadence [steps/min]	3.5	102.71 ± 12.34	101.37 ± 6.42	100.35 ± 8.89
	4	107.97 ± 7.70	104.15 ± 7.38	105.46 ± 7.79
Step Length [m]	3.5	0.57 ± 0.03	0.57 ± 0.05 (5**)	0.57 ± 0.04 (4**)
	4	0.60 ± 0.03	0.61 ± 0.03 (3**)	0.60 ± 0.05 (5**)
Single Support Phase [% stride]	3.5	64.20 ± 5.43	63.45 ± 3.14* (8**)	64.13 ± 3.42 (8**)
	4	66.36 ± 4.25	63.93 ± 4.32* (8**)	64.55 ± 3.11* (9**)

Median ± IQR among subjects. The step length was normalized to the subject height. The information in parentheses indicate the number of subjects that experienced a significant reduction in the respective gait parameter.

** $p < 0.05$, and * $p < 0.15$ indicate a significant reduction in comparison with normal walking.

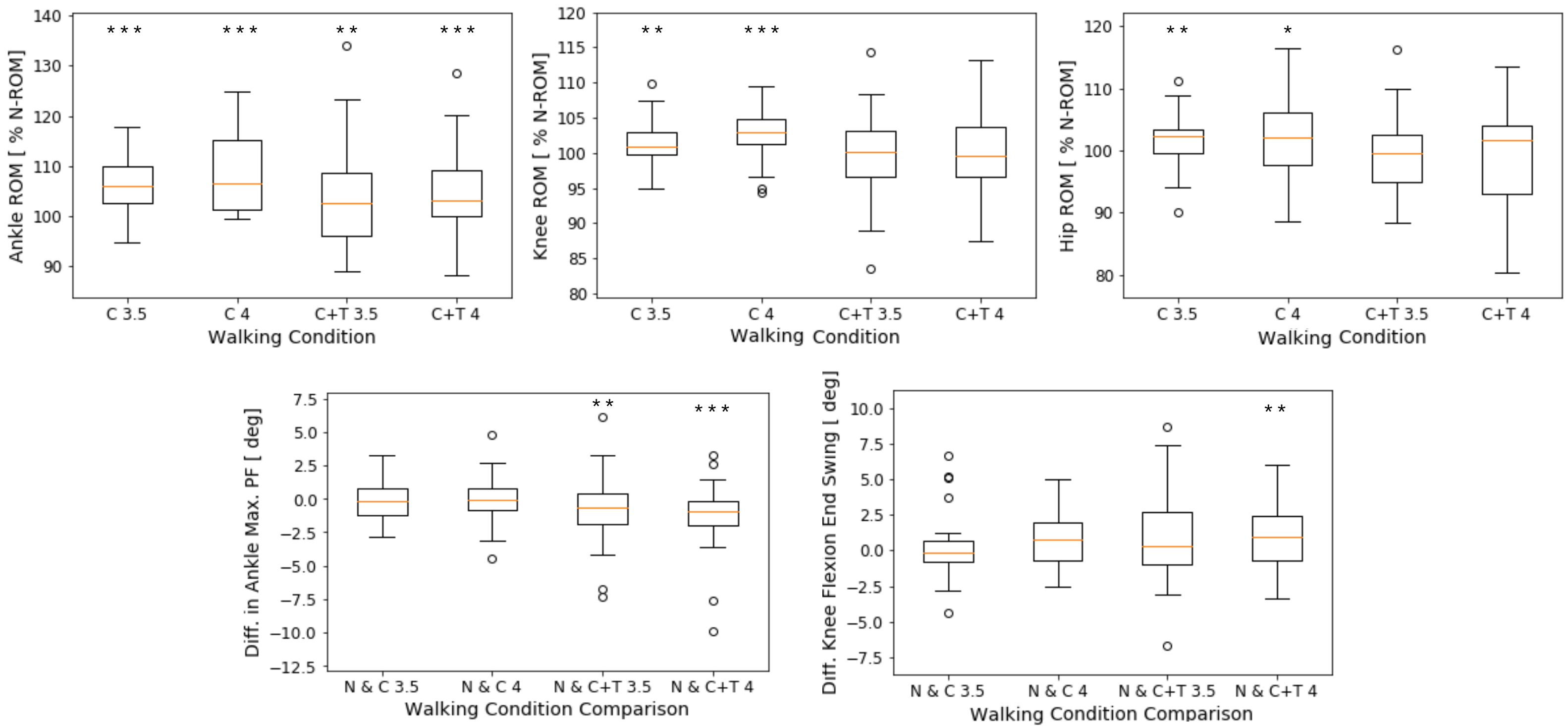


Fig.2.5 **Joint motion between subjects.** (Upper row) Joint ROM during C-restriction (C) and CT-restriction (C+T), at 3.5 km/h and 4 km/h walking speeds, expressed as a percentage of the ROM during normal walking. (Lower row) Difference between restricted (C, C+T) and normal walking (N) in the maximum ankle plantar flexion and knee flexion at the end of the swing phase. A negative value indicates that the respective parameter decreases during restricted walking. *** $p < 0.01$, ** $p < 0.05$, and * $p < 0.1$ indicate significant differences with normal walking. The edges of the box plot correspond to the 25-th and 75-th percentiles, and the single points (outliers) correspond to the respective parameter values that were distant from the general tendency.

The ankle ROM increased significantly during C-restriction ($p < 0.01$), and less significantly during CT-restriction ($p < 0.05$). The knee ROM increased significantly during C-restriction ($p < 0.05$), and tended to be lower than that in normal walking during CT-restriction, but this was not significant. The hip ROM increased significantly during C-restriction ($p < 0.1$); during CT-restriction, it reduced, but was higher than that during normal walking (not significantly). The ankle maximum plantar flexion at the end of the swing phase reduced significantly for both C-restriction ($p < 0.13$ at 3.5 km/h) and CT-restriction ($p < 0.05$). The knee flexion at the end of the swing phase tended to be higher for both C- and CT-restrictions at 4 km/h, and it was significantly higher during CT-restriction ($p < 0.05$).

Considering each subject independently at both walking speeds, the following number of subjects exhibited the following, at C- and CT-restriction, respectively: Three and seven subjects exhibited lower ankle ROM, six and ten subjects exhibited lower knee ROM, nine and seven subjects exhibited higher hip ROM, nine and six exhibited a lower plantar flexion peak, and ten and nine subjects exhibited higher knee flexion at the end of the swing phase.

MTC reduction

Table 2.2 lists the mean, SD, median and IQR of the MTC central tendency between subjects, for each leg, during normal and restricted walking, at both walking speeds. The MTC central tendency is displayed as a percentage of the body height, and its value corresponding to the average height (1.72 m) of the subjects is listed.

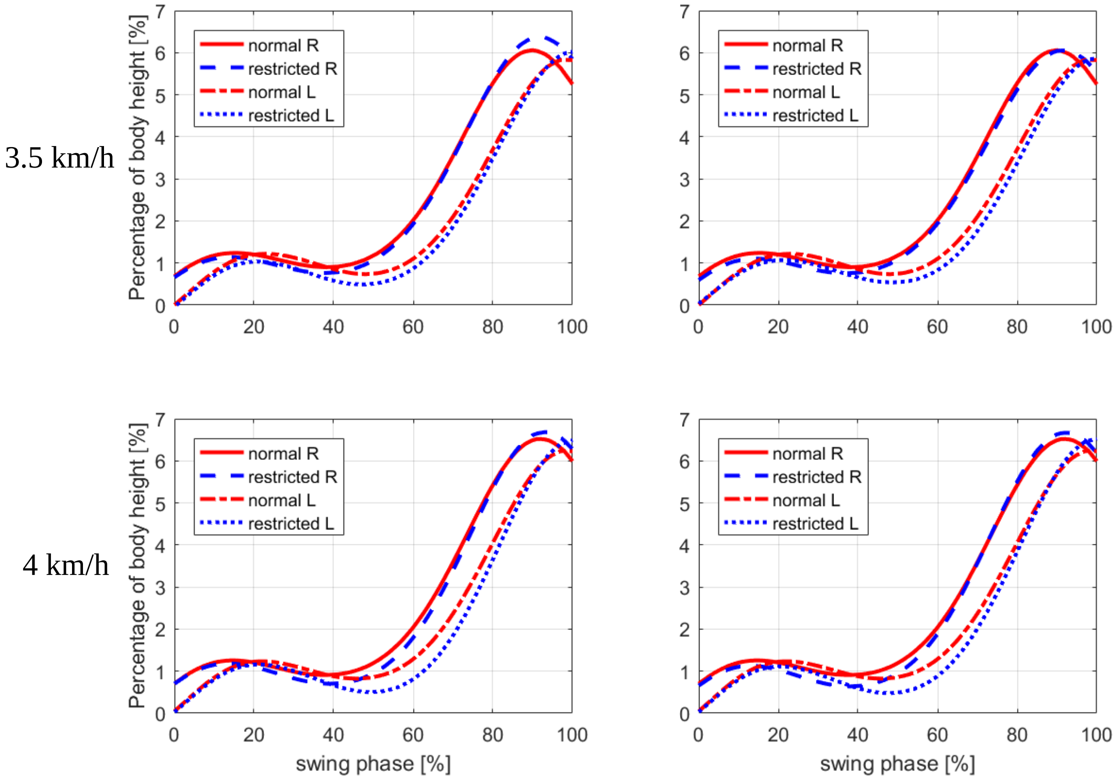
In agreement with the reported MTC range in young adults [47], the clearance during normal walking corresponded to a range of 10-20 mm. The MTC median value in the restricted cases was within the MTC range of the elderly reported by previous studies (e.g., 7.1 mm [46] and 12.9 mm [50]). The MTC in restricted walking was statistically proven to be lower than that in normal walking, which agrees with the relationship between the elderly and the youth [46, 74, 114].

Table 2.2 **MTC central tendency between subjects**

Walking speed [km/h]	Leg	Normal walking		C-restriction		CT-restriction	
		% o.b.h.	Average subject [mm]	% o.b.h.	Average subject [mm]	% o.b.h.	Average subject [mm]
3.5	Left	0.64±0.26 (0.62±0.36)	11.1±4.4	0.40±0.31*** (0.38±0.41)	6.9±5.4	0.47±0.41* (0.35±0.62)	8.2±6.9
	Right	0.71±0.25 (0.68±0.36)	12.3±4.3	0.59±0.18*** (0.59±0.23)	10.1±3.1	0.59±0.29* (0.57±0.41)	10.1±4.9
4	Left	0.60±0.30 (0.64±0.53)	10.3±5.1	0.44±0.30** (0.34±0.34)	7.5±5.2	0.41±0.30** (0.36±0.55)	7.0±5.1
	Right	0.65±0.18 (0.63±0.21)	11.1±3.1	0.56±0.16** (0.58±0.19)	9.6±2.8	0.47±0.17*** (0.42±0.32)	8.0±2.9

Mean ± SD (Median ± IQR). Central tendency of the subjects' MTC expressed in percentage of the body height (% o.b.h.), and in that corresponding to the average subjects' height (1.72 m). *** $p < 0.01$, ** $p < 0.05$, and * $p < 0.1$ indicate the significant differences compared to normal walking.

The average pattern of the mean toe clearance of all the subjects is shown in Figure 2.6. The MTC, which is the lowest point observed in the middle-swing phase, was located between 40-50% of the swing phase, in agreement with the study of Mills et al. [49]. For each subject, the location of the MTC, during restricted walking, did not differ from that in normal walking. At the MTC instant, the swing leg and the trunk were located ahead of the support leg. The MTC, under the C- and CT-restriction cases, was lower than that during normal walking for both legs and walking speeds.



(a) Normal walking and C-restriction (b) Normal walking and CT-restriction

Fig.2.6 **Average toe clearance during the swing phase.** The MTC is the lowest point observed in the middle-swing phase, and is located between 40-50% of the swing phase. The MTC in restricted walking (C- and CT-restriction cases) is lower than that in normal walking for the right (R) and left (L) legs, at both 3.5 km/h and 4 km/h walking speeds.

Another measure of the reduction in MTC is the number of toe contacts to the ground in the middle-swing phase. Table 2.3 lists the number of toe-contact events of each subject in terms of the percentage of the total gait cycles. Although toe contacts occurred even during normal walking, the occurrences under the C-restriction case were higher than those during normal walking, and even higher, under the CT-restriction case, for all subjects.

The number of toe contacts with the ground was more than twice in the C-restriction case, and more than five times in the CT-restriction case, in comparison with normal walking. The ease of reducing the MTC is subject dependent, and as shown in Table 2.3, three subjects (B, D and E) experienced toe-contacts in almost every stride for both restriction conditions, and subject-G experienced the same frequency of toe-contacts in the strongest restriction condition, which is the CT-restriction at the walking speed of 4 km/h.

Additionally, the dominant leg of the subjects (right leg) tended to contact the ground less frequently than the other leg, and that the number of contacts tended to be higher at the faster speed (4 km/h).

Table 2.3: **Frequency of toe contacts to the ground**

Subject	Walking Speed [km/h]	Normal walking [%]		C-restriction [%]		CT-restriction [%]	
		Left	Right	Left	Right	Left	Right
A	3.5	2.06	3.44	6.60	10.76	10.53	10.18
	4	0.66	3.31	3.83	6.39	9.68	9.35
B	3.5	2.72	4.08	>90	>90	>90	>90
	4	1.15	2.29	>90	>90	>90	>90
C	3.5	0	16.78	10.84	34.97	39.66	64.14
	4	2.85	16.14	10.65	33.14	20.65	37.10
D	3.5	0	1.26	>90	>90	>90	>90
	4	0	2.24	>90	>90	>90	>90

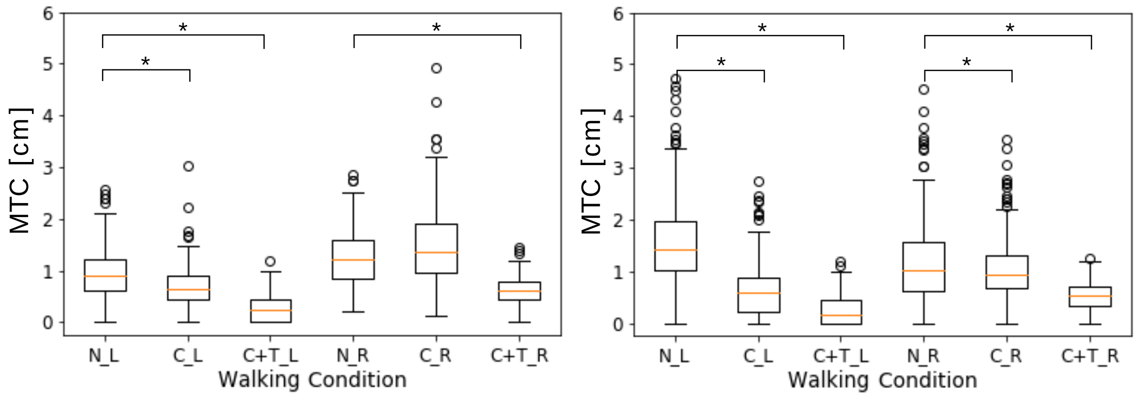
Subject	Walking Speed [km/h]	Normal walking [%]		C-restriction [%]		CT-restriction [%]	
		Left	Right	Left	Right	Left	Right
E	3.5	1.92	9.29	>90	>90	>90	>90
	4	1.87	6.85	>90	>90	>90	>90
F	3.5	0	1.20	0.86	3.45	0.77	6.51
	4	0	0	0	0.63	18.93	11.24
G	3.5	1.04	6.92	1.69	1.69	8.06	9.52
	4	2.32	2.98	4.89	4.23	>90	>90
H	3.5	0.76	1.90	1.10	5.86	3.70	5.56
	4	0	0.66	0.95	7.94	6.51	12.70
I	3.5	0	0	0	8.94	2.27	19.42
	4	0	0.61	0.91	17.93	9.57	36.23
J	3.5	1.86	4.83	2.41	10.69	1.98	6.93
	4	0	0.66	0.65	7.14	2.68	11.74

Toe-ground contacts during middle swing phase. The number of contacts is expressed as a percentage of the total gait cycles.

MTC distribution

The distribution of the MTC was not Gaussian in all the walking conditions (normal and restricted) and walking speeds. As previously reported [46], the MTC is positively skewed and leptokurtic.

Figure 2.7 shows the main characteristics of the MTC distribution of a representative subject, because a similar tendency was observed in all the subjects. In addition to the lower MTC median, the variability (IQR) of the MTC in restricted walking tended to be lower than that in normal walking.



(a) MTC of left and right legs at 3.5 km/h

(b) MTC of left and right legs at 4 km/h

Fig.2.7 **MTC median and variability (IQR) of a representative subject.** The general tendency of the MTC of the subjects during restricted walking (C = C-restriction case, C + T = CT-restriction case) included a lower MTC median and IQR, compared to normal walking (N), for both right (R) and left (L) legs. * $p < 0.01$ indicates a significant lower MTC median compared to normal walking. The edges of the box plot correspond to the 25-th and 75-th percentiles, and the single points (outliers) correspond to the MTC values that were distant from the general tendency.

Table 2.4 shows the IQR of the MTC averaged among the subjects. The IQR corresponding to CT-restriction was lower than that in the C-restriction, and the IQR is higher at a speed of 4 km/h. The MTC IQR tends to be higher in the dominant leg, during normal and restricted walking.

Table 2.4 **MTC variability (IQR) among subjects.**

Walking Speed [km/h]	Leg	Normal walking [mm]	C-restriction [mm]	CT-restriction [mm]
3.5	Left	5.16 ± 1.35	4.98 ± 1.75	4.43 ± 1.33 *
	Right	6.04 ± 1.31	5.01 ± 1.81 *	4.75 ± 2.01 *
4	Left	5.84 ± 2.49	5.37 ± 1.63	4.48 ± 1.27 *
	Right	6.13 ± 1.90	4.94 ± 1.44 *	4.92 ± 2.21 *

$*p < 0.1$ indicate a significant reduction in the IQR in comparison with normal walking.

2.5 Discussion

Effect of MARTT in the reduction of the MTC

The reduction in MTC can be described by the changes in the joint angles, resulting from the restriction of the expansion range of the muscles. From the point of view of the MTC instant, the subjects exhibited the following: when the muscles at the calf alone were restricted, the MTC was reduced due to a lower dorsal flexion of the ankle (the maximum flexion decreased by 1.7° at 3.5 km/h and by 0.6° at 4 km/h, on an average). Some subjects additionally exhibited lower hip flexion, which enhanced the MTC reduction. When the muscles at both the calf and thigh were restricted, the MTC was mainly reduced by a lower hip flexion. The maximum hip flexion decreased by 2.6° at 3.5 km/h and by 3.0° at 4 km/h, on average. In few subjects, the restriction at the thigh also caused a reduction in the knee flexion. Thus, the principal restriction effect of the C-restriction was the reduction in the ankle flexion, and that of the CT-restriction was the reduction in the hip flexion at the MTC instant.

Furthermore, can MARTT reproduce the changes in the joints' motion, in the rest of the gait cycle, which compensate for the low MTC, and reproduce certain gait characteristics caused by muscle weakness, reported in elderly walking? As mentioned in the study of Prince et al. [115], considering the complete gait cycle, the elderly exhibit the following changes in the joint motion: higher hip ROM, higher knee ROM, and higher knee flexion at the end of the swing phase, which are actions for a safer walking. Additionally, lower ankle ROM, and a lower ankle plantar flexion peak were also reported, which are signs of muscle weakness in the elderly. These changes, other than a reduced ankle ROM, were also found in most subjects, when MARTT was implemented, as indicated in the Results section and Figure 2.5. The ankle ROM was higher after MARTT was applied because of the higher ankle dorsal flexion at terminal stance, which is due to the weakness of the gastrocnemius muscle that cannot deaccelerate the ankle dorsal flexion after mid-stance.

This result suggests that by using MARTT, it is feasible to reproduce the MTC and the compensation motions characterizing the walking of the elderly, in young adults. Chapter 3 offers a detailed examination of the compensation motions seen in young adults with MARTT and a comparison with the compensation motions reported for the elderly.

Effect of the walking speed

MARTT reduced the MTC in both the C- and CT-restriction conditions, regardless of the walking speed. However, the combination of CT-restriction and a walking speed of 4 km/h (faster speed) achieved the highest reduction in MTC, as shown in Table 2.2.

Moreover, as it is natural for the human body to compensate restricted motions, certain subjects exhibited higher hip and knee flexion during the swing phase, when the ankle flexion was reduced by C-restriction. For these subjects, this compensation was reduced or eliminated at a walking speed of 4 km/h or when the thigh was restricted.

Similarities and differences between MARTT-restricted youth and the elderly

MARTT can simulate a higher number of MTC values near zero clearance, as observed in the elderly, which can potentially lead to a fall. This increment in the frequency of low MTC values was reflected in a considerable number of toe contacts with the ground, for all the subjects, as listed in Table 2.3. Therefore, reproducing the higher ground-contact risk of the elderly is feasible using MARTT.

The difference in the gait motion between the elderly and the young adults restricted by MARTT was observed in the MTC variability. The elderly present a higher MTC variability in comparison with young adults. However, during restricted walking with MARTT, the variability of the MTC of the subjects was lower than that in normal walking due to the limitation of the full range of flexion of the hip and ankle joints by applying MARTT, which prevented the exceeding of a certain maximum flexion value, in the subjects. As a result, the sporadic occurrences of high MTC values commonly observed in the elderly during walking [46] due to their loss of control of muscle contraction, were not simulated. These sporadic high MTC values are the reason of the higher MTC variability of the elderly; however, they do not represent any risk of falling.

When both restriction cases are compared, the number of toe ground contacts was higher, when both the calf and thigh muscles were restricted. Thus, the restriction at the thigh can be mainly useful for reducing the MTC of subjects, who are sufficiently skillful to overcome the restriction at the calf muscles.

Chapter 3

Biomechanical analysis of gait compensation motions due to muscle weakness

This chapter focuses on the study of the gait compensation motions exhibited by young subjects walking with MARTT in the restriction conditions: C- and CT- restriction (refer to Chapter 2 for details on MARTT and the restriction conditions).

The compensation action of a joint is considered as the change in its flexion or extension range in comparison to its normal range of motion, that works to overcome the effects of muscle weakness. Muscle weakness refers to the inability of the muscle to reach its original muscle activity that is required to carry out its respective tasks. In the context of this study, muscle weakness is the result of natural aging or the restriction applied by MARTT.

In order to identify which compensation strategies are implemented by the subjects, the compensation motions along the complete gait cycle were observed and classified. The differences in the compensation motions across the walking speeds, muscle restriction conditions, and subjects were analyzed.

Figure 3.1 depicts the muscles and muscles groups that are referred to in this study.

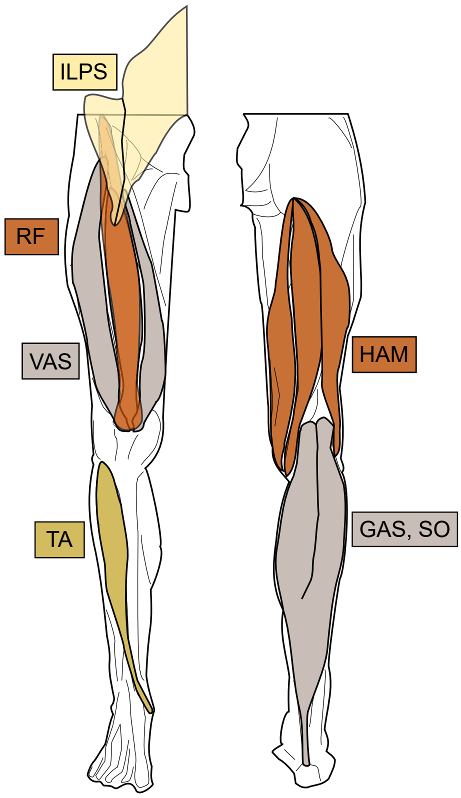


Fig.3.1 **Location and abbreviations of muscles referred in this study.** ILPS: iliopsoas muscle, RF: rectus femoris muscle, VAS: vastii muscle group, HAM: hamstring muscle group, GAS: gastrocnemius muscle, SO: soleus muscle, TA: tibialis anterior.

3.1 Hypothesis

In the previous chapter, MARTT was validated to be a technique that can reduce the MTC of young adults. Since lowering the MTC increases the risk to trip, as observed in Chapter 2 in the increment of the foot-ground contacts of the subjects, the human body is prone to execute compensation motions that lower such risks in order to maintain a safe walking. Such phenomenon is observed in the elderly, due to their natural gait deterioration, and is described in the following section.

Thus, it was hypothesized that young adults whose lower-limb muscles are restricted by MARTT would show compensation motions during walking such the ones seen as effect of muscle weakness that come with aging. In order to verify this hypothesis, the identified compensation motions were validated based on the reported effects of muscle weakness in the gait simulations of Van der Krogt [116]. Furthermore, similarities to the compensation motions reported in the elderly were considered.

3.2 Compensation motions due to muscle weakness

With the reduction of the lower-limb muscle mass, the gait suffers from several impairments because the loss of mass causes changes in the activity magnitude of the affected muscles and their periods of action along the gait cycle. For instance, a shorter range of activity of the GAS muscle has been reported in elderly with recurrent falls [117], and a lower activity of the same muscle has been found in the elderly during the late stance phase [118]. This last might explain the elderly less-powered ankle push-off [119]. Aniansson et al. [120] reported a reduction of 25–35% in the muscle strength of the leg extensors in 70–80-year-old men.

In addition, a peroneal nerve disfunction, that is a peripheral neuropathy commonly seen in the elderly, causes a lower activity or paralysis of the tibialis anterior (TA) muscle that in turn leads to foot drops (a gait abnormality where the forefoot drops when getting closer to the ground) [121].

Thus, the motion pattern of the lower-limb joints deviates from the normal gait pattern, partially because of the effects of muscle weakness, and partially as a result of unconscious compensation measures that the human takes to minimize the effects of muscle weakness and maintain safe ambulation [122]. For instance, an increased hip power has been reported in the elderly during the stance phase and has been suggested as a compensation for their lower-powered push-off in the late stance phase [92]. Owing to a weak TA muscle during the swing phase, the elderly exert a reduced ankle dorsal flexion; however, to maintain a safe ground clearance, the knee and/or hip joints compensate with a higher flexion [123].

According to the simulations of Van der Krogt [116] regarding the effects of muscle weakness on a lower limb, a weakened muscle can compensate for itself by exerting higher activity (as measured by electromyography), or can be compensated for by other muscles. For instance, the main leg muscle groups can compensate in the following manner:

(a) The weakness of the GAS and soleus (SO) muscle groups increases the activity of the BFS muscle, decreases the activity of the TA muscle and increases the activity of the ILPS muscle. Both the GAS and BFS muscles are responsible for knee flexion in the swing phase; thus, the BFS compensates for the GAS function. Additionally, the GAS and the SO muscles are responsible for plantar flexion and for controlling the rate of dorsal flexion during stance phase. The disability to exert plantar flexion and to control the dorsal flexion is compensated by lowering the activity of the TA muscle that is responsible for the ankle dorsal flexion. The ILPS muscle compensates for the push-off power that the GAS cannot provide at the end of the stance phase.

(b) The weakness of the hamstring (HAM) and VAS muscles, which are responsible for knee flexion/hip extension and knee extension, respectively, increases the activity of these same muscle groups, and decreases the activity of the iliopsoas (ILPS) muscle, responsible for hip flexion. In this case, the HAM and VAS muscles compensate for themselves, and the ILPS muscle compensates by balancing the hip flexion.

(c) The weakness of the RF muscle, which is responsible for knee extension and hip flexion, increases the activity of the ILPS and VAS muscles, and decreases the activity of the BFS muscle. The VAS and BFS muscles compensate for the knee extension: the VAS by taking over the knee extension fuction and the BFS by reducing the knee flexion. The ILPS muscle compensates for the hip flexion function.

3.3 Analysis of the joints' motion as a result of MARTT

3.3.1 Data processing

The first minute of the recorded gait motion of each trial was excluded from the analysis, as it was considered as an adaptation time to the experimental conditions. Then, the position data of all markers were smoothed using a 6 Hz Butterworth filter. The timing of the gait events (heel contact and toe off) were determined according to the ground reaction force observed at the foot soles. Every gait cycle was determined as the gait motion occurring between two consecutive heel contacts of the same leg.

The angles of the lower-limb joints in every gait cycle were calculated with the human motion analysis software SIMM; this software obtains the position of markers placed along the body, as recorded by the motion capture system. The joint motion patterns were normalized to a 0–100% gait cycle, and were separated into the following gait sections: loading response (first 10%), middle stance (10–30%), terminal stance (30–50%), pre-swing (50–62%), initial swing (62–75%), middle swing (75–85%), and terminal swing (85–100%) [124]. Each section of the gait cycle is shown in Figure 3.2.

Furthermore, the joint angles in every gait section were analyzed across subjects to determine the characteristics (differences) of the restricted gait (C-restriction and CT-restriction) in comparison to the normal gait. Several of the found characteristics occurred at the beginning or end of the mentioned gait sections. In that case,the first or last 5% of the corresponding section, respectively, was taken for further statistical analysis. In the section where a characteristic was identified, the respective joint angle was averaged for every subject and walking condition.

The statistical significance within subjects of the characteristics in comparison to the normal gait was tested using a two-tailed paired t-test and compensated with the Bonferroni method to counteract the multiple comparisons effect.

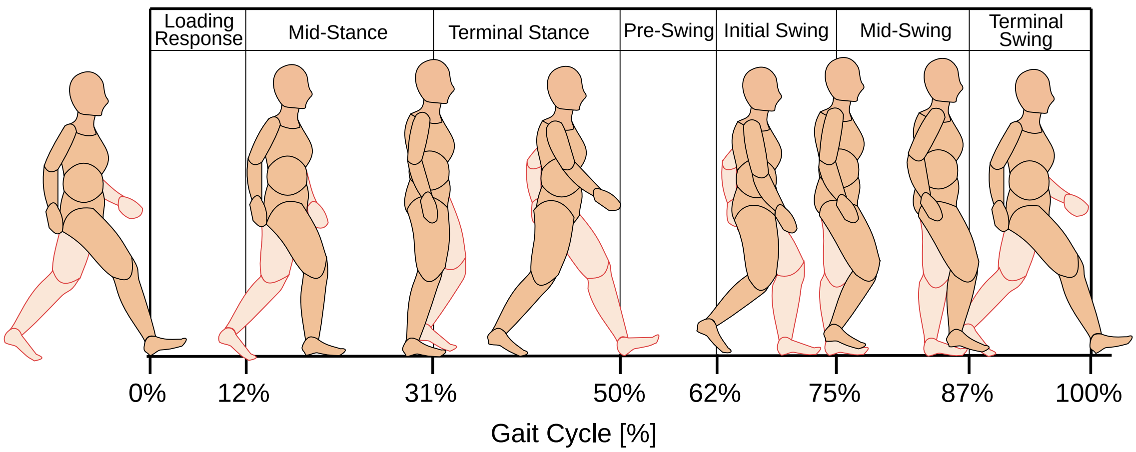


Fig.3.2 **Human gait cycle.** Sections of the gait cycle that are referred to in this study.

3.3.2 Joint motion characteristics

The joint motion characteristics observed in every restriction type and walking speed will be described, along with the number of subjects that showed each characteristic and the significance evaluated in every subject (as specified with p-values). Figure 3.3 shows the joint motion for the C- and CT-restrictions of the left leg of a subject; this represents the motions seen in the majority of the subjects.

Joint motion in C-restriction

In the C-restriction condition, MARTT belts are applied at the calf only.

Ankle joint:

In the ankle joint, the majority of subjects exhibited (a) higher plantar flexion at the beginning of the loading response phase, (b) higher dorsal flexion in the terminal stance phase, and (c) lower dorsal flexion during the swing phase (see these characteristics in Figure 3.3a). At the speed of 3.5 km/h, (a) seven ($p < 0.01$), (b) eight ($p < 0.01$), and (c) seven ($p < 0.01$) subjects presented these characteristics, respectively. Furthermore, at the speed of 4 km/h, (a) six (five with $p < 0.01$ and one with $p < 0.05$), (b) nine ($p < 0.01$), and (c) seven (six with $p < 0.01$ and one with $p < 0.05$) subjects showed the same characteristics.

Knee joint:

In the knee joint, all subjects showed higher knee flexion, mainly in the (a) loading response and (b) terminal stance phases, and (c) from the initial to middle swing phase (see Figure 3.3c). At the speed of 3.5 km/h, (a) eight (six with $p < 0.01$ and two with $p < 0.05$), (b) seven ($p < 0.01$), and (c) seven ($p < 0.01$) subjects presented this characteristic, respectively. Furthermore, at the speed of 4 km/h, (a) eight ($p < 0.01$), (b) seven ($p < 0.01$), and (c) seven ($p < 0.01$) subjects showed this characteristic.

Hip joint:

In the hip joint, the subjects mainly exhibited (a) higher hip extension in the middle stance (see Figure 3.3e), (b) higher hip extension in the terminal stance (see Figure 3.3e), (c) higher hip flexion in the initial swing and (d) higher hip flexion in middle swing. At a walking speed of 3.5 km/h, (a) four ($p < 0.01$), (b) six (five with $p < 0.01$ and one with $p < 0.05$), (c) five (four with $p < 0.01$ and one with $p < 0.05$), and (d) six ($p < 0.01$) subjects presented these characteristics, respectively. At the speed of 4 km/h, the same trend was observed in (a) ten (eight with $p < 0.01$ and two with $p < 0.05$), (b) nine ($p < 0.01$), (c) seven ($p < 0.01$), and (d) eight ($p < 0.01$) subjects, respectively.

Joint motion in CT-restriction

In the CT-restriction condition, MARTT belts are applied at the calf and thigh.

Ankle joint:

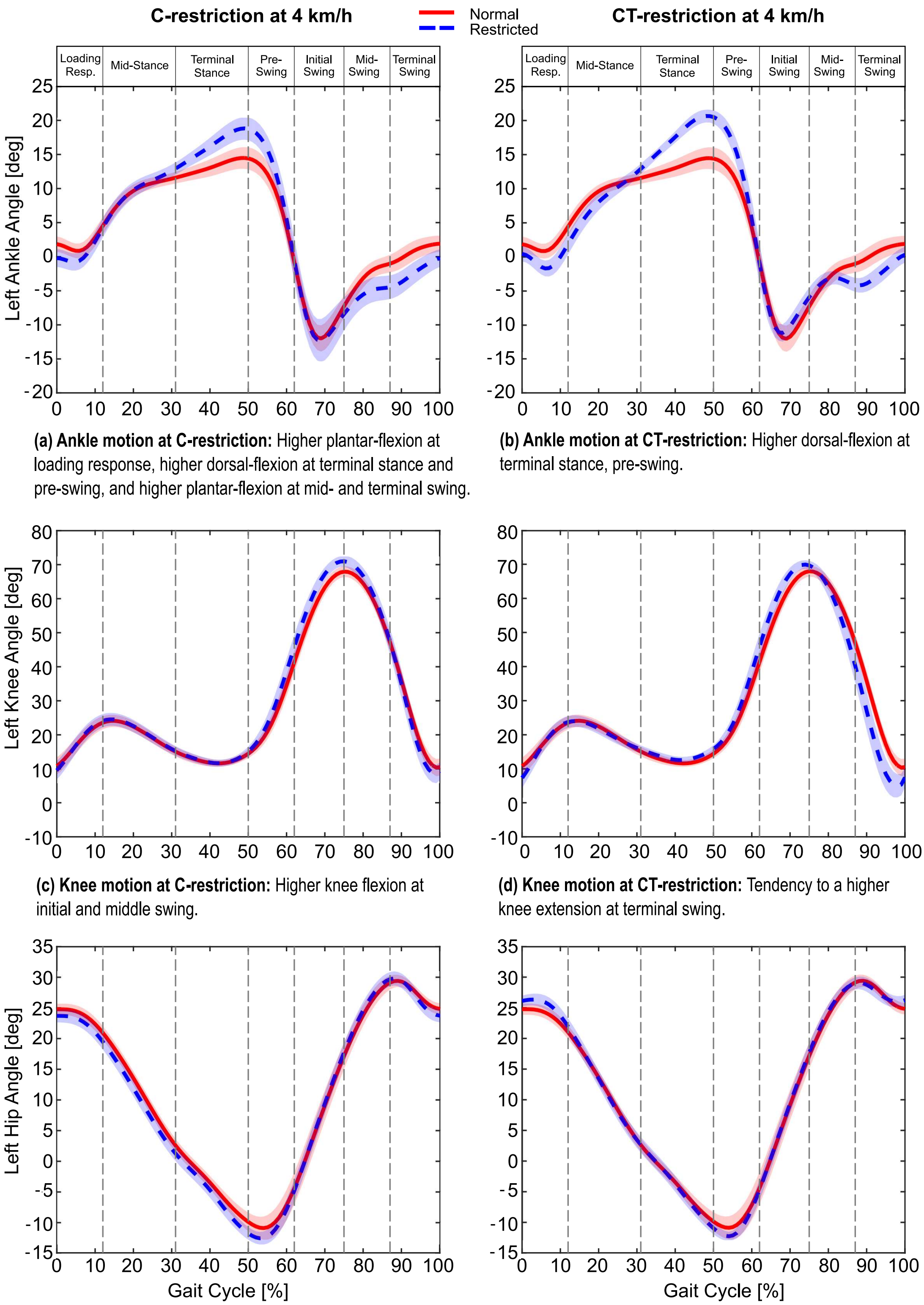
In the ankle joint, the subjects mainly presented higher dorsal flexion in the (a) terminal stance (see Figure 3.3b) and (b) initial and terminal swing. At the speed of 3.5 km/h, (a) eight ($p < 0.01$) and (b) seven ($p < 0.01$) subjects presented higher dorsal flexion, respectively. At the speed of 4 km/h, the same trend appeared in (a) ten ($p < 0.01$) and (b) eight ($p < 0.01$) subjects, respectively. The higher dorsal flexion in the terminal stance phase was also seen in the C-restriction in the same eight subjects at the speed of 3.5 km/h and in nine subjects at 4 km/h.

Knee joint:

In the knee joint, seven (six with $p < 0.01$ and one with $p < 0.05$) subjects presented a higher extended knee in the terminal swing phase (see Figure 3.3d) at the walking speed of 3.5 km/h, and nine ($p < 0.01$) subjects at the speed of 4 km/h.

Hip joint:

In the hip joint, the majority of the subjects exhibited (a) higher hip extension at the beginning of the pre-swing phase (see Figure 3.3f), (b) lower hip flexion during the initial and middle swing phases, and (c) lower hip flexion in the terminal swing phase. At the walking speed of 3.5 km/h, (a) eight (seven with $p < 0.01$ and one with $p < 0.05$), (b) seven ($p < 0.01$), and (c) ten subjects (nine with $p < 0.01$ and one with $p < 0.05$) presented these compensations, respectively. At the walking speed of 4 km/h, these compensation motions appeared in (a) nine ($p < 0.01$), (b) nine (eight with $p < 0.01$ and one with $p < 0.05$), and (c) seven ($p < 0.01$) subjects, respectively.



(a) Ankle motion at C-restriction: Higher plantar-flexion at loading response, higher dorsal-flexion at terminal stance and pre-swing, and higher plantar-flexion at mid- and terminal swing.

(b) Ankle motion at CT-restriction: Higher dorsal-flexion at terminal stance, pre-swing.

(c) Knee motion at C-restriction: Higher knee flexion at initial and middle swing.

(d) Knee motion at CT-restriction: Tendency to a higher knee extension at terminal swing.

(e) Hip motion at C-restriction: Tendency to a higher hip extension from loading response until the end of terminal stance.

(f) Hip motion at CT-restriction: Tendency to a higher hip extension at pre-swing.

Fig.3.3 **Joint motion of a representative subject.** Typical ankle, knee and hip joints motion observed in the majority of the subjects.

3.4 Classification of compensation motions and prioritization of gait characteristics

3.4.1 Compensation strategies

In the following, the mentioned (previous section) changes in the motion patterns of each joint as a result of MARTT are analyzed taking into account the compensation strategies reported in the simulations of Van der Krogt, i.e., those reported as a consequence of the weakness of certain muscles or muscle groups. With this analysis,the compensation motions were classified.

Strategies in C-restriction

In the C-restriction condition, the mainly restricted muscles are the GAS and TA. Figure 3.4 summarizes the compensation actions that were observed in this restriction type, in the respective section of the gait cycle.

Ankle joint:

According to Van der Krogt's simulations, when the TA and GAS muscles are weakened, their activity level reduces as the weakness increases. Additionally, when the GAS and SO muscle groups are weakened, the TA muscle activity decreases. The compensation motion observed in our study as shown in Figure 3.3a agrees with the decay in the TA muscle activity during the loading response and swing phases, where, in a normal gait, the TA muscle is known to have its most prominent activity in generating a dorsal-flexion moment [124].

The higher plantar flexion during the loading response phase observed in our experiment suggests that the movement of the foot toward plantar flexion after heel contact is not decelerated by a weak TA muscle, as it occurs in the normal gait. Thus, the ankle reaches a higher plantar flexion than normal before a sufficient dorsal-flexion moment is created at the ankle that moves the tibia forward over the articular surface of the talus (known as ankle rollover). During the swing phase, the ankle exhibited a lower dorsal flexion, resulting in a lower foot clearance.

The weakness of the GAS muscle was noticeable in the majority of subjects from the mid-stance to terminal stance, when the GAS in the normal gait is most active in controlling the dorsal flexion of the ankle before toe off [124]. At the end of the rollover, the ankle reached a higher dorsal-flexion angle before initiating plantar flexion to prepare for toe off. This agrees with the concept of a weakened GAS muscle that could not decelerate the dorsal flexion of the ankle as normal.

Knee joint:

An increased knee flexion from the loading response to mid-stance was observed in the majority of subjects (see Figure 3.3c), that suggests a weakness in the VAS muscles. In the normal gait, these muscles prevent knee hyperflexion in the loading response phase, and extend the knee until the middle of mid-stance [124].

Moreover, the observed higher knee flexion in the terminal stance and initial swing suggests compensation from the BFS muscle as reported in the simulations of Van der Krogt, i.e., as a compensation for the weak GAS and SO muscles. The higher activity of the BFS muscle in a terminal stance could help prevent knee hyperextension, and the higher activity in the initial swing could compensate for the lower dorsal flexion of the ankle and ensure sufficient foot clearance.

Hip joint:

Van der Krogt's simulations reported increased activity in the HAM muscles as compensation for the weakness of the GAS muscle. In agreement with this compensation, a higher hip extension was observed from the loading response to the middle of the mid-stance (see Figure 3.3e), where in normal gait the semimembranosus and semitendinosus muscles (HAM muscles) are active to extend the hip. In addition, an increased hip extension was observed at the end of the terminal stance phase, which suggests a weakness in the tensor fascia lata that limits the hip extension before the hip starts to flex in the pre-swing in the normal gait.

Moreover, the increased hip flexion observed in the initial swing suggests increased activity in the ILPS muscle, also seen in Van der Krogt's simulations, as a compensation for the GAS muscle weakness. This could be an action to compensate for the lack of the hip flexion moment that is normally generated in pre-swing and is usually enough to drive the hip passively during the initial swing phase [124]. As a result, the extra hip flexion would support the above-mentioned compensation of the knee, and contribute to the foot clearance.

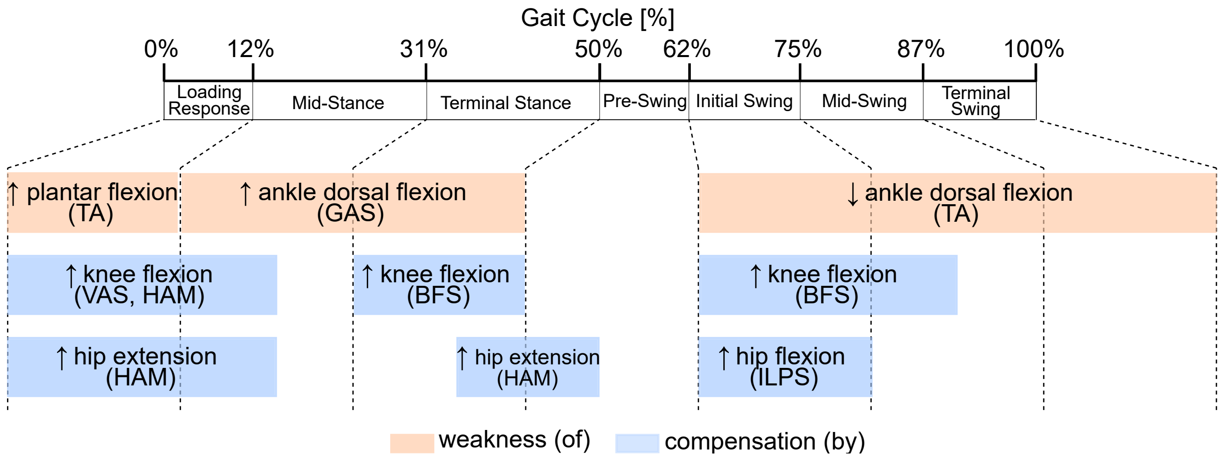


Fig.3.4 **Compensation strategies in C-restriction.** The orange boxes show the changes in joint' motion that are the result of the weakness of a specific muscle or muscle group. The blue boxes show the changes in joint' motion that are the result of compensation actions of a specific muscle or muscle group.

Strategies in CT-restriction

In the CT-restriction condition, the RF, HAM, and VAS muscles are restricted in addition to the muscles that are restricted in the C-restriction condition. Figure 3.5 summarizes the compensation strategies that were observed in this restriction type, in the respective section of the gait cycle.

Ankle joint:

An increased ankle dorsal flexion was observed during the swing phase (see Figure 3.3b), which suggests an increased activity of the TA muscle, that is a compensation for the restricted HAM muscles in Van der Krogt's simulations. This action might serve to safely clear the ground, as the knee and hip may not be able to compensate with a higher flexion as they did in C-restriction. The higher ankle dorsal flexion observed in the terminal stance suggests a weak GAS muscle, as also seen in the C-restriction, but in this case, the flexion is higher. The weak GAS is one of the reported effects from restricting the RF muscle in Van der Krogt's simulations.

Knee joint:

A higher knee extension was observed in the terminal swing phase (see Figure 3.3d). This might be caused by the decreased activity in the HAM muscles and/or increased activity in the VAS muscles, as reported by Van der Krogt during the swing phase when the VAS muscles are weakened. During the swing phase in the normal gait, the HAM muscles modulate the rate of knee extension, and the VAS muscles act to ensure a complete extension, so as to prepare for the abrupt weight transfer at heel contact. Weak HAM muscles would fail to regulate the knee extension, and a higher activated VAS would produce a higher knee extension, and compensate for the lower hip flexion that was observed in the majority of subjects at terminal swing (section below). This would otherwise significantly reduce the step length.

Hip joint:

A lower hip flexion was observed at pre-swing (see Figure 3.3f), suggesting weakness in the RF and gracilis (GRA) muscles. These contribute to flexing the hip, to prepare for the swing phase in the normal gait. This partially agrees with Van der Krogt's results, as in that study, a higher activity in the RF muscle during pre-swing and a lower activity in the GRA muscles along the gait cycle were reported as effects of the RF muscle restriction.

The observed reduced hip flexion during the initial and middle swing phases also suggests lower activity in the GRA muscle, which normally supports the flexion at this time. This would mean that the hip cannot compensate to clear the ground, as it did in C-restriction. However, as mentioned in the compensations for the ankle, the TA muscle compensates to clear the ground.

The observed higher extended hip in the terminal swing phase is the result of the previously mentioned lack of the flexion action of the GRA muscle. As a result of the weak hip flexion during the swing phase, the step length would be shorter than normal.

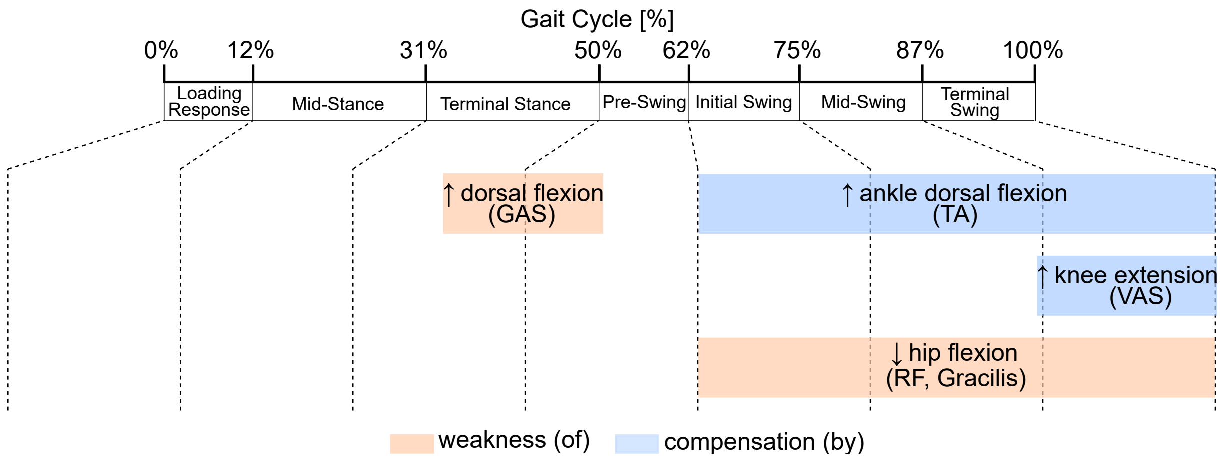


Fig.3.5 **Compensation strategies in CT-restriction.** The orange boxes show the changes in joint' motion that are the result of the weakness of a specific muscle or muscle group. The blue boxes show the changes in joint' motion that are the result of compensation actions of a specific muscle or muscle group.

In brief, in the C-restriction case, the subjects compensated for the restricted ankle motion with the non-restricted hip and/or knee joints. In the CT-restriction case, where the ankle, knee, and hip joints were restricted, the majority of subjects compensated for the ankle joint with the joint itself. With the analysis above, I have come to the classification of the compensation strategies, of young adults walking with MARTT, shown in Figure 3.6. This compensation behavior agrees with the results reported in the gait simulations of Van der Krogt [116], who studied the effects of muscle weakness on human gait. The simulations suggested that a weakened muscle might react by decreasing or increasing its own activity, or the activity of other muscles. In other words, the weakened muscle can compensate for itself or can be compensated by other muscles.

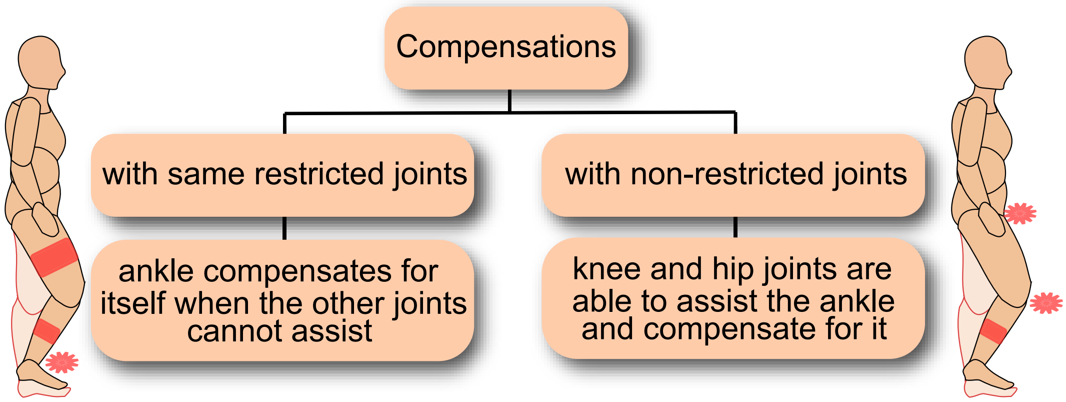


Fig.3.6 **Compensation strategies classification.** Identified categories of the compensation strategies observed in young adults walking with MARTT.

3.4.2 Prioritization gait characteristics

Based on the observations in the subsection above:

Prioritization in C-restriction: When the muscles at the calf area were restricted (C-restriction), the weakness of the ankle in reaching the normal dorsal flexion angle during the swing phase was compensated for by a higher knee and hip flexion, showing a priority in compensating for foot drop. The knee also compensated to protect itself from hyperextension in the terminal stance phase. The hip compensated for itself during the initial swing phase by flexing more to create the necessary moment, normally generated in the terminal stance phase, to advance the leg forward. The hip also compensated for the hyperflexion of the knee in the loading response phase by extending more. In this muscle restriction condition, the weakness of the joint was always compensated for within the kinematic chain, either by the joint itself or by others.

Prioritization in CT-restriction: When the muscles at the calf and thigh area were restricted (CT-restriction), the ankle compensated for itself during the swing phase with a higher dorsal flexion, thereby preventing foot drop. The knee and hip were not able to aid in this task as they did in C-restriction. The hip did not reach the normal flexion range during the swing phase, which reduced the step length. However, some subjects compensated for the higher extended knee in the terminal swing phase to keep the step length. In this muscle restriction condition, the active compensation chain of the C-restriction was not seen. However, the subjects' gait motion appeared to always prioritize safe ambulation, as also seen in the elderly.

By comparing the compensation actions manifested in the C- and CT-restrictions, a more dynamic compensation activity can be seen in C-restriction; this is because the ankle, knee, and hip joints (kinematic chain) are able to compensate for each other, and in some cases even themselves. One of the reasons for this is the severity of the restriction. As the knee and hip joints are minimally affected in C-restriction, they can supply what the overall kinematic chain needs. However, in CT-restriction, the muscle restriction affects the complete kinematic chain; therefore, the body has to prioritize what to compensate. It was found that the prioritization includes the prevention of foot drop, knee hyperextension in the terminal stance phase, and knee hyperflexion in the loading response phase, as well as the preservation of the step length. This suggests that the priority is to adapt the gait patterns for safe walking, as also seen in the elderly [74]. Moreover, the evaluation of the significance of every joint motion characteristic in every subject showed that the significance increased at a faster walking speed of 4 km/h in both C- and CT-restriction.

3.5 Discussion

3.5.1 Joint motion compensation and weakness: similarities and differences with the elderly

The resultant joint motion of the young adults walking with MARTT in the C-restriction condition agreed with several compensation motions and weaknesses seen in the healthy elderly. For instance, the young adults presented a lower ankle plantar flexion in the terminal stance at the time of push-off, as seen in the elderly [91, 119]. In addition, a lower dorsal flexion of the ankle during the swing phase was seen in the young adults, which is also commonly known in the elderly as foot drop [114].

Öberg et al. [96] reported the following joint motion changes with aging: increment in knee flexion in the middle stance phase, reduction of knee flexion in the swing phase, and an increment in hip flexion in the mid-swing phase. In agreement, the young adults with C-restriction showed these motion changes, except for the lower knee flexion during the swing phase. However, the higher knee flexion during the swing phase (as seen in these young adults) has also been found in other studies of the elderly [114, 115] as a safety measure to compensate for foot drop.

3.5.2 Limitations

The subject sample that participated in this study is relatively small, and it is possible that there exist other compensation patterns not seen in our analysis. Thus, it is necessary to observe the compensation motions in a much larger subject sample to investigate the universal trend of compensation strategies. Furthermore, a muscle activity analysis is missing to validate what has been observed in this study regarding which muscles are active or weak at the moment the described compensation strategies occur.

To investigate the relationship between the restriction of certain muscle groups and compensation motions, MARTT was used to cause the muscle restriction artificially in young subjects. However, the congruence between the compensation motions that result from this artificial restriction and the ones that result from natural muscle weakness have to be further analyzed and validated.

This study has concentrated on the compensation strategies at the lower limb. However, it is also important to analyze the compensation strategies at the upper body and find the similarities and differences with the trunk motions of the elderly.

Chapter 4

Overall discussion

4.1 Reproduction of the elderly gait characteristics with MARTT in young adults

The general topic of this doctoral project was the investigation of the feasibility to reproduce with MARTT age-related gait characteristics of the healthy elderly in young adults. More specifically, this project focused on the following aims:

Aim 1: Compare the MTC central tendency, variability and distribution of young male adults, whose lower limbs have been restricted with MARTT, with the respective values that have been reported in literature about the elderly.

Aim 2: Validate MARTT as a technique that is able to reduce the MTC of the young to the values of the elderly without causing strange walking patterns in the young that do not correspond to the natural human gait cycle.

Aim 3: Compare the spatio-temporal gait parameters of the young adults walking with MARTT with the age-related changes in these parameters reported for the elderly.

Aim 4: Determine the similarities between the gait compensation motions observed in the young adults walking with MARTT, and the age-related compensations reported for the elderly in literature that are a result of muscle weakness.

The content of this section summarizes the principal findings in regard to these aims.

4.1.1 Reproduction of the MTC of the elderly

In Chapter 2, the toe clearance of ten male young adults was analyzed when MARTT was applied to restrict their muscles at the lower limb. The restriction was applied in two different manners: (a) only restricting the shank, and (b) restricting both the shank and the thigh. The young individuals walked in two different walking speeds, and the dependance of the toe clearance to the walking speed was considered.

The results of this study showed that MARTT reduced the MTC of the young adults to the MTC values reported in the elderly. In other words, the median value of the MTC of the young adults reduced to values about 7 to 10 mm, that are within the range of the elderly MTC reported till date in literature [46, 50]. In regard to the distribution of the MTC of the young adults, their MTC during normal and restricted walking was positively skewed and leptokurtic, as it is also for the elderly [46]. This means that MARTT did not impair the toe clearance pattern of the young adults. In regard to the variability of the MTC, the young adults restricted with MARTT showed a lower MTC variability in comparison to their natural MTC variability. This disagrees with the MTC variability of the elderly, for in literature it is shown that the elderly present a higher MTC variability than the young, ought to sporadic high MTC values that are the result of the lost of control of their muscle activation. However, such high MTC values, when talking about tripping or falling, do not represent any risk. Since the general purpose of using MARTT is to reproduce in the young characteristics of the elderly gait that represent a risk in the elderly safety, the reproduction of the mentioned high variability of the elderly MTC is not in the scope of the purpose of MARTT.

As a consequence of lowering the MTC of the young adults, they presented a significant higher number of toe-ground contacts. The toe-ground contacts increased by at least twice, when the shank muscles were restricted, and by five times, when both the shank and thigh muscles were restricted. Such toe-ground contacts are common in the elderly and represent a high risk of tripping and falling down. Being able to reproduce such dangerous circumstances of the elderly in the young, would be of great help to understand how to support the elderly to overcome danger by experimenting in the young with MARTT instead of the elderly. This application of MARTT will be discuss in the following section of this chapter.

The reduction of the MTC in the young adults was achieved by different reactions of the lower-limb joints, that were dependent on the restricted muscles. The restriction of the shank muscles, specially the TA, was meant to lower the ROM of the ankle joint, and in turn lower the MTC. In fact, when MARTT was applied at the shank the dorsal flexion range of the ankle did reduce, but the subjects also showed a higher knee and hip flexion angles that tried to overcome the reduction of the MTC. It was interesting to observe such unconscious and immediate reactions of the young adults, that elderly people also develop throughout the aging process. The restriction of the thigh muscles was meant to reduce such compensations for the MTC, and in fact, this restriction prevented the compensation from the knee and hip, but in this case a compensation from the ankle joint, that increased its ROM, was observed. Due to the compensation triggered at the ankle joint, in this type of restriction condition (shank and thigh muscles restricted), the MTC of the young adults was reduced thanks to a lower hip flexion.

The walking speed appeared to influence the ability of the subjects to compensate for the MTC. At the fastest speed, the compensations for the MTC reduced and with it the value of the MTC. The combination of the restriction of both the shank and thigh muscles and the fastest walking speed achieved the lowest MTC value of the subjects. The young subjects showed a higher MTC variability at the fastest walking speed in their natural gait, and also in their restricted gait for both restriction conditions and both right (dominant leg) and left legs.

4.1.2 Similarities with the compensation strategies of the elderly

As already discussed in this section, the reduction of the MTC of the young adult subjects triggered compensation motions at the lower-limb joints. Since MARTT restricted the lower-limb muscles during the complete gait cycle, it would be very probable to encounter compensation motions not only during the swing phase at the time of the MTC, but also during the stance phase. The reason of this assumption is that there might be several parts of the gait cycle that represent a risk, and are therefore unconsciously prioritized by the Central Nervous System, as the MTC is, which triggers compensation motions as measures to maintain a safe gait.

In Chapter 3, the complete gait cycle of the young adults was analyzed to determine how they compensate the muscle restriction applied by MARTT. The compensation strategies observed when only the shank muscles were restricted agreed with the ones reported in literature on the elderly. For instance, the subjects presented an increment of knee and hip flexion during initial swing phase that agrees with the elderly [114]. For both the elderly and the young subjects with MARTT, these flexion increments would compensate for: first, the push-off power that could not be generated at the end of the stance phase because of the weak GAS muscle; and second, the low MTC caused by the weak TA muscle. Additionally, an increment of knee flexion during middle stance has been reported for the elderly [96]. In agreement, the young subjects showed also the mentioned higher knee flexion, along with, a higher hip extension during the loading response and middle stance. A higher flexed knee and extended hip at the loading response and the middle stance phase could compensate for a safer landing at heel contact and higher stability, to prevent stumbling, during early stance.

In addition, when the shank muscles were restricted, the young subjects showed a higher knee flexion during terminal stance. To the knowledge of the author of this manuscript, this compensation has not been reported for the elderly. However, computational simulations on the compensations that result from the weakness of the GAS and SO muscles [116] found that the BFS muscle compensates, which would result in the higher knee flexion at the terminal stance. This compensation of the BFS muscle can protect the knee from hyperextension at the end of the stance phase. The young subjects also showed a higher hip extension at the end of terminal stance. This change in the hip motion has not been reported for the elderly, and would be important to have it in consideration on further studies on the elderly gait, in order to understand if this motion is caused by a weak muscle or muscle group or if it is a compensation strategy for the low push-off power at the end of stance phase.

When the muscles at the shank and thigh were restricted, the young subjects mainly showed a compensation with the ankle joint during the swing phase. This action expresses how important it is to be able to safely clear the ground during the swing phase. In other words, making sure that there is enough toe clearance is one of the main gait priorities. It is interesting to notice that when only the muscles at the shank were restricted, the knee and hip joints compensated for the ankle; however, when the muscles at both the shank and thigh were restricted, the knee and hip joints were not able to compensate and the ankle joint had to achieve it for itself during the time where it is most needed, which is the MTC time. In literature, such compensation of the ankle during the swing phase has not been reported for the elderly. One reason can be that the elderly subjects that participate in experimental studies do not have such severe muscle weakness that affect their knees and hips as the muscle weakness that MARTT applied to the young subjects when the muscles at both the shank and thigh were restricted. Moreover, the young subjects at this restriction type showed a higher knee extension at the terminal swing. This action can be a measure to preserve the step length and could explain why most of the young subjects did not show a significant change in their step length after MARTT restriction was applied (see Table 2.1).

A thorough study of the elderly gait is necessary to determine if the compensation motions here found in the young subjects, not reported till date in the literature about the elderly, agree or disagree with the elderly compensation strategies. Such study on the elderly is challenging because according to their level of muscle weakness, experimenting on the elderly can expose them to several dangers, that have to be first mitigated.

4.2 Application of MARTT

MARTT was developed and investigated mainly to be utilized to study the elderly gait. The main advantage of implementing MARTT is that in such studies young adults take the place of the elderly, preventing any danger that those studies otherwise would represent for the elderly. Thus, studies in risky scenarios are more possible to be carried out with the help of MARTT.

This section offers a discussion about how MARTT could be of use in the endeavor of understanding the elderly gait and finding more precise assistant techniques for the elderly needs.

4.2.1 Further investigation of the elderly gait

Risky scenarios in which it is important to study the elderly gait include different kind of terrains like: slopes, stairs, terrains with obstacles and irregular terrains. Asking the elderly to repetitively walk in such terrains would represent a high risk to trip and fall. Even though appropriate protective gears like a harness and joint protections are used, the elderly are still exposed to pain and exhaustion; and therefore, ethical principles would limit the participation of the elderly. In such cases, MARTT could be of use; mainly for the study of the risk to trip or fall in the different terrains due to a low MTC. As it will be discussed in the following section about the limitations of MARTT, MARTT can only reproduce in the young certain physical characteristics of the healthy elderly that are the result of muscle weakness. Therefore, MARTT could be used to further the study of the elderly gait only when the studies are narrowed to the capabilities of MARTT. Along Chapter 2 and Chapter 3, MARTT was validated to reproduce the MTC values of the elderly, and equal compensation motions were found. In addition, MARTT reduced the single support phase of the young subjects and increased the toe contacts to the ground as see in the elderly.

A recent study used MARTT to investigate which muscles at the shank are of main importance in the regulation of the level of the MTC in level ground walking [125]. The lower limb muscles of young subjects where restricted with MARTT and surface electromyography was used to monitor the shank muscles. This is an example of studies that are merely dependent on the physical effect of the MTC, where MARTT could be of use. This study could be continued in other terrains to determine the prevalence of shank muscles in regulating the MTC. As also suggested by the author of this study, knowing the muscles and factors that play a critical role in the maintenance of the MTC, care givers could have a clear focus on deciding the rehabilitation or training plans for the elderly.

Moreover, it is also important to investigate how the human body compensates muscle weakness while walking in the mentioned terrains. As the compensation strategies in level ground walking were studied in Chapter 3, MARTT could be used to study the compensation strategies in the other type of terrains. Such study would elucidate the gait characteristics that the human body prioritizes to maintain when certain muscle groups are weak and cannot accomplish their natural tasks. As Winter et al. [74] and Pirker et al. [122] reported in their studies, compensation motions are meant to compensate for muscle weakness with the final goal of maintaining a safe ambulation. Therefore, knowing which gait characteristics are of main importance to warranty a safe ambulation would give clear goals to the development of assistive methods for the elderly gait. How the knowledge gathered with MARTT could aid in finding out such assistive methods will be discussed in more detail at the end of this section.

4.2.2 Combination of MARTT with other aging suits

MARTT could be combined with existing aging suits or also known as silver simulators like the ”Third Age Suit” from the Ford Motor Company, the ”AGNES” from MIT [100], and the ”GERT Suit” [101] (see details about this sort of suits in Chapter 1) to investigate how the gait characteristics of the elderly reproduced by MARTT in young adults are affected by impairments like the narrowing of the visual field, the loss of hearing, joint stiffness, restrictions in the head mobility, between others. The reduction of the MTC and the compensations motions seen in the elderly gait might be a result of a compound of factors that include not only muscle weakness, but also sensory loses and other motor impairments that come with aging. Using MARTT in young adults makes it possible to isolate the sensory impairments that normally afflict the healthy elderly from the physical ones and with the help of the existing aging suits other impairments can be brought to the studies according to what is the focus of the investigation. This method could serve to inquire into the case-effect relation of all the factors that combined result in the known impairments of the healthy elderly.

Another application of combining the existing aging suits with MARTT would be to enhance the understanding of the elderly sensory and motor impairments by young adults, for example professionals like care givers and engineers, and medical students. Such application has been the motivation of the development of the current aging suits, and as also emphasized by the authors of these aging suits, the understanding of the elderly impairments and needs would improve the services the professionals provide to the elderly, and improve the development or adequacy of cars, sidewalks, etc, that will be reflected in the safety and comfort of the elderly.

4.2.3 Identification of assistance requirements for the elderly

The studies that have been already described in this chapter can be carried out with MARTT in the young with no or a few experimental restrictions in comparison to the amount of restrictions that ethical committees would impose to experiments with the elderly. In addition, thanks to the mental and physical strength of the young, the studies could be carried out with more continuity. These two mentioned factors constitute great advantages when talking about the development of assistive devices and finding out effective methods to support the elderly gait. For now, MARTT could be used to find out effective methods to maintain the MTC within safe values and reduce the compensations motions in level ground walking. As it is discussed in the following section about the limitations of MARTT, further validation of MARTT in other terrains than level ground is needed to be able to use this technique in such terrains.

The investigation with MARTT about the gait characteristics that are prioritized while walking in level ground, described in Chapter 3, might serve as a guidance about what the assistive devices should focus on in order to achieve an efficient and effective support for the elderly gait, and to diminish the compensation motions that they perform. Warrantying the coverage of the gait characteristics of highest priority might considerably reduce the energy the elderly need to walk, and reduce the deterioration rate of the muscles and joints they normally overload to perform the compensations motions.

Young subjects restricted by MARTT could walk the assistive devices, such as exoskeletons, and these devices could be programmed and tuned over the course of the experimental studies in a repetitive manner. These experiments in the young could be of great use to determine the sensory devices that are most useful to detect the key gait characteristics, also to find out the most appropriate location for those devices, and to determine an effective range of joint torques and the right timing to apply them. In this manner, the assistive devices can reach an acceptable level of maturity before the tests in the elderly start.

4.3 Limitations of MARTT and further improvements

Ought to the fact that MARTT can reproduce just a part of the healthy elderly gait characteristics, MARTT implies several limitations for the studies that implements this technique because evidently the elderly comprises a collection of physical afflictions and intrinsic characteristics that cannot be completely replaced by the young and any aging simulation technique. In addition, the studies on MARTT have been carried out in laboratory setups that differ with the day-to-day living environmental conditions of the elderly.

In this section, the limitations of implementing MARTT will be discussed, as well as the limitations of the experimental procedures that were used in the validation of MARTT as a technique to reproduce the MTC of the elderly in the young, described in Chapter 2, and the procedures used in the identification of the compensation motions, described in Chapter 3.

4.3.1 Experimental justifications and limitations

The studies carried out till date on MARTT have just included male young adults as subjects. As it is evident in the studies of healthy elderly men and women, the gait characteristics can vary according to the gender because of the difference in their physical constitution. Thus, the study of MARTT in female young adults will be pending to be conducted, mainly to determine the adequate restriction force that can lower the MTC in the range of the MTC values of healthy elderly women. Additionally, the male subject population in the studies of MARTT was entirely Japanese. Therefore, MARTT has to also be evaluated in other populations to validate in them its effectiveness.

Moreover, the studies of MARTT have been carried out on a treadmill for two reasons; first because a significant amount of gait cycles from each subject was needed, and second because a specific walking speed was needed to be set according to our experimental protocol. Even though a treadmill offers certain conveniences, its effect on the gathered gait data should be kept in mind. Several studies in literature [126, 127] have found minor differences between walking on a treadmill and overground, that account for joint moments, muscle activation patterns and joint powers. Ought to the fact that those differences were found to be comparable to the variability of those gait parameters, these studies have validated that walking on a treadmill is similar to overground walking; and therefore they have stated that a treadmill can be used for gait analysis.

The restriction that MARTT belts apply to the young subjects' lower limbs implies blood flow occlusion. As mentioned in Chapter 2, in section "MARTT Technique", blood flow occlusion in moderation has not negative effect in the subjects health; on the contrary, it is even used by athletes to improve endurance. In moderation means that the restriction pressure applied by MARTT must be regulated to a safe and comfortable value. This is why after investigating the appropriate pressure, and taking into account what studies on blood flow occlusion have reported (160-230 mmHg [111, 112]), the pressure that MARTT applies was regulated to 180 mmHg. Moderation also means that the restriction force should be applied for a specific amount of time in order to avoid exhaustion and pain. For this reason, the period of time the young subjects walk with MARTT is limited. For the case of our studies on MARTT, this time was limited to 6 min, but could have been longer because any of the young subjects reported exhaustion or pain.

The number of subjects in which MARTT has been validated is relatively small (10 subjects), even when the gait characteristics of about 250 strides for each subject were analyzed. This amount of data was significant to investigate the feasibility of reproducing the gait characteristics of the elderly in the young. However, a larger amount of subjects is needed to validated the compensations strategies and their classification studied in Chapter 3. In addition, electromyography data of the young with MARTT has not been studied in the context of this doctoral thesis. Such study is of great importance to validate the observations about which muscles were weak and which compensated in the different gait phases described in Chapter 3.

Since MARTT can merely reproduce certain physical gait characteristics of the healthy elderly in the young, sensory and other motor impairments are excluded from the studies. This fact constitutes from one side a limitation of the technique, but it could also be a benefit, depending on the study goals. It is evidently a limitation when the studies focus on understanding in a global manner the principal gait patterns that characterize the majority of elderly individuals. It is a benefit when including all the physical and mental afflictions of the elderly makes it challenging to comprehend their gait and find out clear support and rehabilitation plans for specific impairments.

4.3.2 Future directions

The future investigations about MARTT should cover two main directions: one consists on the improvement on the technique to warranty its reproducibility and effectiveness when used by other researchers, and the second consists on validating MARTT in other terrains and subject population and the found compensation strategies.

Regarding the first direction, in order to warranty the effectiveness of MARTT, the restriction force applied by MARTT belts has to be monitored and the researcher has to make sure the force keeps invariant and is equal in both legs. The force sensors that were embedded in MARTT belts for the studies that were carried out in this doctoral project, and also for the recent study of Perera et al. [125], that implemented MARTT, were FSR (Force-Sensing Resistor) sensors. An investigation to find out which sensors give the most reliable measurements for the needs of MARTT is important to be carried out in the future. Another improvement for MARTT consists on defining the exact characteristics the construction of MARTT belts should follow in terms of width and material. The results of these studies will give important guidelines that will ease the implementation of MARTT by other researchers.

In regard to the second direction, MARTT should be studied in female subjects to mainly determine the necessary restriction force that can reproduce the specific gait characteristics of the elderly that have been validated for Japanese male subjects in the studies of this doctoral thesis. Additionally, other terrains than overground should be incorporated in the future studies of MARTT to validate the performance of the technique. Moreover, future tests should include electromyography data of the young subjects walking with MARTT and compare it with the muscle activity of the elderly in order to drive further conclusions about the similarities on the compensations strategies of the elderly and the young with MARTT.

Moreover, the study of Chapter 3 did not cover the investigation of the compensation strategies of the trunk and the investigation of the joints motion in other than the sagittal plane. The elderly commonly present a forward tilted trunk that lowers their center of mass providing them more stability when walking. Finding out if such compensations of the trunk also occur in young subjects restricted by MARTT could be useful to pinpoint the causes that could serve as hits to further understand the elderly gait.

Chapter 5

Conclusions

5.1 Project findings

- MARTT was validated as a method to reproduce the MTC central tendency and distribution of the elderly in young adults. The MTC median value of the young subjects was significantly reduced to values lower than 10.1 mm, which agrees with the MTC values found in the elderly. The distribution of the MTC of the young walking with MARTT was positively skewed and leptokurtic as it is characteristic for the healthy young and the elderly.

- The MTC variability of the young subjects walking with MARTT was measured in terms of the IQR of the MTC, and it was found that MARTT reduced the MTC variability in discrepancy with the MTC variability of the elderly, that has been reported to have a tendency to increase.

- The reduction of the MTC in the young subjects increased the toe-ground contacts by at least twice, when the shank muscles were restricted, and by five times, when both the shank and thigh muscles were restricted. This indicates an increase in the risk of tripping of the young adults, that is one of the major causes of falling in the elderly.

- There was a tendency in the young subjects to contact the ground less frequently with the dominant leg (right leg), and the number of contacts tended to be higher at the faster speed (4 km/h).

- MARTT reduced the single support phase of the young subjects, and some showed also a significant lower step length, while no significant changes were found on the cadence.

- MARTT reduced the MTC of the young subjects regardless the walking speed, and the lowest MTC was found in the combination of the fastest walking speed with the strongest MARTT restriction (CT-restriction).

- In the C-restriction case, the subjects compensated for the restricted ankle motion with the non-restricted hip and/or knee joints. In the CT-restriction case, where the ankle, knee, and hip joints were restricted, the majority of subjects compensated for the ankle joint with the joint itself. This compensation behavior agrees with the report from computational muscle weakness simulations gait, that suggest that the weakness of a muscle or muscle group is compensated by other muscles in the kinematic chain or by the muscle itself.

- Similarities were found between the weakness and compensation strategies found in C-restriction and the ones reported in literature for the elderly: lower ankle plantar flexion in the terminal stance at the time of push-off, lower dorsal flexion of the ankle during the swing phase, higher knee flexion in the middle stance phase, higher knee flexion during the swing phase, and higher hip flexion at initial mid-swing phase.

- The young subjects walking with MARTT prioritized the compensation for foot drop, knee hyperextension in the terminal stance phase, and knee hyperflexion in the loading response phase, and to maintain the step length.

5.2 Future directions

The future directions of this project were discussed at the end of Chapter 4. The following lists them in brief.

- Further research on MARTT should address the improvement of MARTT belts in terms of force sensors, size and materials, so that the technique could be reproduced by the research community with ease.

- The subject pool should be expanded to validate the compensation strategies and classification we have suggested in this doctoral project.

- An upper-body compensation study should be carried out, given that this project has only covered the study of the compensation strategies at the lower limb.

- The observations on the compensation strategies found in this project should be validated in a study that includes electromyography data of the most relevant muscle groups of the lower limb.

- The compensation motions should be studied in other terrains and in laboratory-free environments, given that this project has only given insights on level-ground walking.

Appendix A

Subject gait parameters

A.1 Abbreviations

The following are the abbreviations used in the tables:

BMI	Body mass index
C-restriction	Calf restriction applied with MARTT
CT-restriction	Calf and thigh restriction applied with MARTT
IQR	Inter-quartile range
MTC	Minimum toe clearance
MARTT	Muscle Activity Restriction Taping Technique
ROM	Range of motion
SD	Standard deviation

A.2 Subjects

Table A.1 **Subjects characteristics**

Subject	Height [m]	Weight [kg]	BMI [kg/m^2]	Dominant leg
A	1.79	77	24.2	Right
B	1.72	68	23	Right
C	1.75	61.5	20.1	Right
D	1.67	52	18.7	Right
E	1.77	65.7	21	Right
F	1.71	54.2	18.5	Right
G	1.78	74.5	23.6	Right
H	1.66	72.2	26.3	Right
I	1.68	59.8	21.1	Right
J	1.76	54.9	17.7	Right

Characteristics of the subjects that participated in this project

A.3 Spatio-temporal parameters

Table A.2 **Cadence**

Subject	Walking speed [km/h]	Normal walking	C-retriction	CT-restriction
A	3.5	94.78 ± 1.5	93.1 ± 1.42	92.78 ± 1.84
	4	100.1 ± 1.78	101.72 ± 1.48	101.74 ±1.64
B	3.5	110.26 ± 4.48	102.86 ± 2.74	110.82 ± 5.68
	4	97.42 ± 2.8	100.54 ± 3.48	114.24 ± 6.8
C	3.5	104.48 ± 2.5	98.24 ± 2.76	96.58 ± 3.24
	4	104.3 ± 2.12	104.04 ± 4.26	104.06 ± 2.8
D	3.5	106.2 ± 3.14	104.58 ± 2.36	106.36 ± 2.8
	4	121.54 ± 4.4	118.74 ± 2.7	107.44 ± 2.4
E	3.5	97.04 ± 2.32	94.86 ± 2.86	96.42 ± 2.08
	4	107.9 ± 2.9	102.48 ± 2.18	103.04 ± 2.18
F	3.5	111.8 ± 2.82	112.26 ± 3.12	115.96 ± 4.6
	4	118.7 ± 4.48	120.06 ± 4.18	121.08 ± 4.44
G	3.5	100.94 ± 1.5	100.54 ± 1.32	101.52 ± 2.66
	4	108.04 ± 1.78	104.26 ± 1.74	106.62 ± 1.64
H	3.5	90.14 ± 7.66	103.64 ± 2.18	99.18 ± 4.46
	4	108.16 ± 4.86	105 ± 4.52	104.3 ± 7.14
I	3.5	108.18 ± 3.2	102.2 ± 2.48	103.54 ± 2.06
	4	110.26 ± 2.3	110.72 ± 2.08	112.3 ± 2
J	3.5	93.64 ± 1.66	96.62 ± 2.12	97.32 ± 1.58
	4	101.28 ± 2.02	101.52 ± 1.36	100.26 ± 1.44

Subjects' cadence in steps/min. Mean ± SD

Table A.3 **Single support phase**

Subject	Walking speed [km/h]	Normal walking	C-retriction	CT-restriction
A	3.5	68.78 ± 0.94	68.54 ± 0.95	68.61 ± 1
	4	70.43 ± 0.91	69.61 ± 0.85	69.45 ± 0.87
B	3.5	67.85 ± 1.64	61.81 ± 2.08	63.95 ± 3.99
	4	64.89 ± 2.19	59.97 ± 2.32	64.4 ± 2.93
C	3.5	68.19 ± 3.1	64.96 ± 1.91	65.41 ± 4.79
	4	67.83 ± 1.21	67.05 ± 2.22	67.06 ± 1.95
D	3.5	62.47 ± 1.01	62.72 ± 1.12	64.3 ± 1.13
	4	63.13 ± 0.86	63.22 ± 1.04	64.69 ± 1.18
E	3.5	65.58 ± 1.53	64.22 ± 1.94	64.68 ± 1.79
	4	67.87 ± 1.4	64.63 ± 1.64	66.28 ± 1.39
F	3.5	66.18 ± 1.48	65.18 ± 1.33	62.9 ± 2.17
	4	68.96 ± 2.36	66.72 ± 1.43	65.73 ± 2.14
G	3.5	61.83 ± 0.91	59.69 ± 0.93	61.44 ± 0.98
	4	63.9 ± 1.08	62.56 ± 0.9	62.02 ± 1.13
H	3.5	61.67 ± 5	64.17 ± 3.83	69.74 ± 3.1
	4	68.22 ± 4.35	70.4 ± 3.47	63.15 ± 4.29
I	3.5	62.82 ± 1.61	59.74 ± 1.31	58.75 ± 1.45
	4	61.87 ± 1.21	59.7 ± 1.12	60.79 ± 1.22
J	3.5	61.85 ± 0.92	61.57 ± 1.38	61.06 ± 1.45
	4	63.88 ± 1.41	62.93 ± 1.11	63 ± 0.96

Subjects' single support phase in % of gait cycle. Mean ± SD

Table A.4 **Step length**

Subject	Walking speed [km/h]	Normal walking	C-retriction	CT-restriction
A	3.5	35.29 ± 1.68	35.85 ± 1.68	35.85 ± 1.68
	4	38.66 ± 1.68	37.54 ± 1.68	37.54 ± 1.68
B	3.5	34.30 ± 2.33	36.63 ± 2.91	27.91 ± 5.81
	4	33.72 ± 2.33	32.56 ± 2.33	31.98 ± 5.23
C	3.5	28.60 ± 4.58	32.04 ± 2.29	33.18 ± 5.15
	4	34.32 ± 2.29	34.90 ± 3.43	34.90 ± 2.86
D	3.5	35.37 ± 1.80	34.77 ± 1.80	33.57 ± 1.80
	4	34.77 ± 1.80	35.37 ± 1.80	38.97 ± 1.80
E	3.5	34.52 ± 1.13	34.52 ± 1.70	34.52 ± 1.13
	4	35.09 ± 1.13	36.79 ± 1.13	36.79 ± 1.13
F	3.5	32.69 ± 2.34	32.11 ± 1.75	31.52 ± 2.92
	4	36.19 ± 4.67	34.44 ± 1.75	34.44 ± 3.50
G	3.5	32.64 ± 1.69	32.08 ± 1.13	31.51 ± 1.69
	4	35.45 ± 1.69	36.02 ± 1.69	34.89 ± 1.69
H	3.5	32.61 ± 6.04	30.19 ± 3.62	35.02 ± 2.42
	4	34.42 ± 5.43	37.44 ± 3.62	32.61 ± 4.23
I	3.5	31.51 ± 1.19	33.29 ± 1.19	33.29 ± 1.19
	4	35.08 ± 1.19	35.08 ± 1.19	35.08 ± 1.19
J	3.5	33.52 ± 1.70	32.95 ± 2.27	33.52 ± 1.70
	4	36.36 ± 1.70	36.36 ± 1.70	37.50 ± 1.70

Subjects' step length in % of their body height. Mean ± SD

A.4 Joints range of motion

Table A.5 **Ankle ROM**

Subject	Walking speed [km/h]	Normal walking		C-restriction		CT-restriction	
		Left	Right	Left	Right	Left	Right
A	3.5	27.02	28.54	28.35	29.56	27.69	29.9
	4	29.24	30.05	29.55	30.64	29.1	30.99
B	3.5	31.11	30.58	31.45	32.72	28.1	27.26
	4	32.65	30.4	33.07	33.73	28.79	27.22
C	3.5	27.14	28.52	31.95	32.83	32.35	31.88
	4	32.72	32.52	34.83	33.05	34.2	34.08
D	3.5	27.38	28.39	29.03	32.01	28.21	30.74
	4	26.48	28.14	31.11	35.16	31.81	36.15
E	3.5	25.97	30.81	25.47	32.15	23.88	32.2
	4	25.54	30.23	27.6	34.73	25.74	33.16
F	3.5	24.18	19.52	26.18	20.13	23.05	18.54
	4	24.33	19.49	26.4	20.77	24.4	19.89
G	3.5	37.08	37.04	36.1	39.31	36.4	37.7
	4	36.07	34.66	36.11	40.54	37.27	38.04
H	3.5	24.15	26.73	26.51	29.56	32.36	32.96
	4	25.93	30.75	30.68	35.6	30.47	33.2
I	3.5	33.52	33.22	36.07	38.35	35.47	36.31
	4	36.65	38.28	37.11	38.5	36.07	36.57
J	3.5	31.78	32.34	31.13	30.7	32.23	31.21
	4	32.09	33.38	34.22	33.21	34.96	34.24

Subjects' ankle joint ROM in degrees

Table A.6 **Knee ROM**

Subject	Walking speed [km/h]	Normal walking		C-restriction		CT-restriction	
		Left	Right	Left	Right	Left	Right
A	3.5	59.97	59.75	61.26	61.49	60.46	60.36
	4	62.39	62.93	63.84	63.62	63.22	62.67
B	3.5	63.32	63.55	63.19	60.34	58.48	53.12
	4	61.85	60.5	64.83	61.59	58.11	55.03
C	3.5	60.27	53.93	60.92	56.59	61.95	56.13
	4	64.13	57.48	61.95	56.9	59.93	56.46
D	3.5	59.91	61	61.43	60.54	58.46	57.46
	4	57.57	57.06	62.48	59.58	64.87	64.65
E	3.5	64.52	64.04	64.37	64.46	65.31	63.66
	4	64.97	63.86	66.85	67.12	65.8	65.83
F	3.5	55.86	54.97	54.12	52.7	51.73	48.86
	4	56.22	55.67	53.07	52.85	49.12	50.36
G	3.5	66.8	61.76	67.53	61.83	66.12	60.24
	4	64.5	61.31	68.86	63.46	66.8	60.98
H	3.5	53.78	56.67	56.04	56.85	61.47	61.45
	4	54.07	61.31	59.15	64.36	59.5	61.11
I	3.5	58.94	62.29	64.69	66.99	62.48	65.92
	4	63.56	66.42	64.44	67.92	61.95	65.9
J	3.5	74.5	67.62	74.05	69.61	73.49	69.31
	4	72.6	69.14	74.71	71.19	75.63	71.72

Subjects' knee joint ROM in degrees

Table A.7 **Hip ROM [deg]**

Subject	Walking speed [km/h]	Normal walking		C-restriction		CT-restriction	
		Left	Right	Left	Right	Left	Right
A	3.5	34.57	34.74	35.59	35.74	36.49	36.5
	4	37.19	37.43	36.33	37.39	37.14	38.18
B	3.5	38.01	35.89	37.81	37.17	36.81	36.15
	4	38.34	37.68	44.62	43.64	42.35	39.77
C	3.5	41.18	36.63	38.73	35.29	36.48	35.38
	4	43.75	39.27	40.69	37.21	39.33	37.33
D	3.5	39.36	39.33	40.42	42.78	39.12	43.24
	4	40.29	39.46	42.35	44.38	41.33	43.63
E	3.5	34.86	36.79	35.68	36.76	36.07	36.14
	4	35.76	36.66	37.96	38.92	37.04	38.18
F	3.5	34.23	37.1	38.05	33.39	30.26	33.13
	4	38.1	38.61	38.75	34.21	30.65	35.19
G	3.5	34.71	36.52	35.81	36.79	40.37	33.67
	4	36.35	38.23	39.07	38.22	41.26	35.28
H	3.5	37.27	37.33	37.05	37.99	35.66	38.2
	4	37.91	40.11	37.91	41.15	39.45	35.07
I	3.5	35.71	36.31	37.19	37.74	35.71	36.54
	4	37.18	37.81	38.87	39.24	37.62	38.61
J	3.5	42.87	39.02	42.17	39.96	39.82	38.93
	4	44.18	42.05	41.76	40.86	41.23	40.25

Subjects' hip joint ROM in degrees

A.5 MTC

Table A.8 **MTC**

Subject	Walking speed [km/h]	Normal walking		C-restriction		CT-restriction	
		Left	Right	Left	Right	Left	Right
A	3.5	0.46 ± 0.32	1.2 ± 0.47	0.08 ± 0.22	1.07 ± 0.34	0.4 ± 0.28	1.18 ± 0.31
	4	0.44 ± 0.28	1.19 ± 0.26	0.34 ± 0.25	1.22 ± 0.39	0.57 ± 0.28	1.24 ± 0.29
B	3.5	1.23 ± 3.19	2.11 ± 0.78	0.02 ± 0.08	1.3 ± 0.45	0.21 ± 0.46	1.26 ± 0.9
	4	0.15 ± 0.34	1.71 ± 0.65	0.02 ± 0.08	1.34 ± 0.37	0.21 ± 0.53	1.14 ± 1.04
C	3.5	1.86 ± 1.82	0.77 ± 0.68	0.15 ± 0.5	0.31 ± 0.48	0.02 ± 0.17	0.22 ± 0.33
	4	1.62 ± 1.9	0.73 ± 0.75	0.32 ± 0.59	0.46 ± 0.45	0.02 ± 0.19	0.36 ± 0.4
D	3.5	0.87 ± 0.44	0.93 ± 0.67	0.94 ± 0.57	1.24 ± 0.72	1.45 ± 0.54	1.88 ± 0.78
	4	1.63 ± 0.6	0.82 ± 0.57	1.62 ± 0.87	1.24 ± 0.86	1.29 ± 0.58	0.65 ± 0.77
E	3.5	0.51 ± 0.5	0.71 ± 0.42	0.48 ± 0.44	1.03 ± 0.42	0.35 ± 0.47	0.65 ± 0.43
	4	0.51 ± 0.37	0.76 ± 0.46	0.54 ± 0.47	0.96 ± 0.51	0.4 ± 0.48	0.76 ± 0.39

Subjects' MTC in cm. Median ± IQR

Table A.9 **MTC**

Subject	Walking speed [km/h]	Normal walking		C-restriction		CT-restriction	
		Left	Right	Left	Right	Left	Right
F	3.5	0.91 ± 0.61	1.23 ± 0.73	0.64 ± 0.48	1.36 ± 0.94	0.23 ± 0.42	0.6 ± 0.35
	4	1.44 ± 0.96	1.02 ± 0.96	0.59 ± 0.64	0.95 ± 0.63	0.17 ± 0.44	0.53 ± 0.36
G	3.5	0.87 ± 0.36	1.23 ± 0.47	0.7 ± 0.35	1.31 ± 0.42	0.96 ± 0.43	1.37 ± 0.48
	4	0.85 ± 0.42	1.24 ± 0.51	0.89 ± 0.38	1.06 ± 0.41	0.71 ± 0.35	1.12 ± 0.45
H	3.5	1.35 ± 0.7	1.6 ± 0.78	1.29 ± 0.6	0.95 ± 0.41	2.21 ± 0.5	1.5 ± 0.26
	4	1.12 ± 0.74	1.2 ± 0.74	0.96 ± 0.56	1.08 ± 0.48	1.41 ± 0.5	1.08 ± 0.48
I	3.5	1.5 ± 0.5	1.57 ± 0.5	0.81 ± 0.43	0.72 ± 0.31	0.79 ± 0.47	0.58 ± 0.36
	4	1.01 ± 0.56	0.95 ± 0.47	0.58 ± 0.49	0.51 ± 0.36	0.77 ± 0.55	0.48 ± 0.34
J	3.5	1.54 ± 0.7	0.9 ± 0.54	1.81 ± 0.89	0.87 ± 0.52	1.49 ± 0.69	0.85 ± 0.55
	4	1.53 ± 0.99	1.57 ± 0.76	1.64 ± 0.58	0.82 ± 0.48	1.45 ± 0.58	0.72 ± 0.4

Subjects' MTC in cm. Median ± IQR

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