UNIVERSIDADE FEDERAL DO RIO GRANDE DO SUL
ESCOLA DE EDUCAÇÃO FÍSICA, FISIOTERAPIA E DANÇA
PROGRAMA DE PÓS-GRADUAÇÃO EM CIÊNCIAS DO MOVIMENTO HUMANO

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Functional and coordination determinants of gait in older adults

PORTO ALEGRE
2021

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Thesis submitted in partial fulfillment of the requirements for the degree of Doctor in Human Movement Sciences of the School of Physical Education, Physiotherapy and Dance at the Universidade Federal do Rio Grande do Sul.

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Porto Alegre
2021

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Final concept: approved

December 1, 2021

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CIP - Catalogação na Publicação

Feijo Martins, Valeria
Functional and coordination determinants of gait in older adults / Valeria Feijo Martins. -- 2021.
108 f.
Orientador: Leonardo Alexandre Peyré-Tartaruga.

Coorientador: Andrea Kruger Gonçalves.

Tese (Doutorado) -- Universidade Federal do Rio Grande do Sul, Escola de Educação Física, Programa de Pós-Graduação em Ciências do Movimento Humano, Porto Alegre, BR-RS, 2021.

1. functional and coordination determinants of gait in older adults. 2. older adults. 3. idoso. 4. marcha de idoso. I. Alexandre Peyré-Tartaruga, Leonardo, orient. II. Kruger Gonçalves, Andrea, coorient. III. Título.

Elaborada pelo Sistema de Geração Automática de Ficha Catalográfica da UFRGS com os dados fornecidos pelo(a) autor(a).

I dedicate this work to my parents, sister and fiancé, because they are all by my side supporting me and giving me strength to continue in academic life.

AGRADECIMENTOS

Primeiramente, saliento a minha gratidão e sinceros agradecimentos aos meus amigos e orientadores Leonardo e Andrea. Obrigada por compartilharem seus conhecimentos e transmitir seus ensinamentos, de acreditarem no meu potencial e de me apresentarem a este mundo que é a pesquisa em alto nível. Obrigada pela confiança e pelas oportunidades. Vocês fazem parte das minhas conquistas e da minha trajetória acadêmica.

Agradeço aos membros e colaboradores do grupo LOCOMOTION (Grupo de Pesquisa em Mecânica e Energética da Locomoção Terrestre) e do grupo CELARI (Centro de Estudos de Lazer e Atividade Física do Idoso), que foram parceiros em todos os momentos necessários para a realização desta pesquisa.

Agradeço a minha família, ao meu pai e minha mãe que sempre incentivaram a busca pela educação e conhecimento. A minha irmã por dividir momentos acadêmicos comigo. Ao meu noivo que sempre me apoiou e me incentivou nas minhas decisões e escolhas.

Agradeço aos meus amigos que em vários momentos me acalmaram e me ajudaram a não surtar (risos) e a alinhar os pensamentos e objetivos dessa longa jornada.

Agradeço pelas oportunidades que me foram concedidas ao longo desses anos, como inúmeras viagens a congressos nacionais e internacionais e principalmente, a professora Bárbara Pellegrini do grupo CERISM (*Sport, Montagna e Salute, Università di Verona*) por ter me recebido para realizar um estágio de pesquisa aplicada em Rovereto - Itália.

Muito obrigada a todos!

RESUMO

Os idosos têm padrão de marcha prejudicado, diminuição da velocidade, fraqueza muscular, controle motor prejudicado, coordenação dos membros prejudicada, ângulos e comprimento da passada diminuídos, aumento do tempo de apoio duplo e deficiências posturais. No entanto, ainda não há uma compreensão clara dos efeitos do envelhecimento sobre os principais determinantes da marcha. O objetivo geral desta Tese foi, portanto, examinar os determinantes funcionais e de coordenação da marcha em idosos. Para atingir esse objetivo, três estudos originais foram desenvolvidos: 1) explorar os determinantes físicos do declínio da velocidade da marcha em 187 idosos ativos (72,2±6,8 anos). O desempenho da caminhada foi caracterizado por 2 parâmetros: índice de reabilitação locomotora (IRL) e índice de caminhada (RC). Com o avançar da idade, as velocidades auto selecionadas e máximas diminuíram em até 26%, assim como as variáveis de aptidão física pioraram com o envelhecimento. Encontramos a RC e o equilíbrio corporal como preditores da velocidade máxima, sugerindo que seria necessário diminuir a RC para evitar o desequilíbrio. Reduções excessivas nos parâmetros analisados indicam perda da homeostase na mecânica da marcha. 2) compreender até que ponto os mecanismos de controle do membro superior estão presentes durante a marcha. 20 idosos e 13 jovens caminharam em diferentes velocidades em esteira e foi calculada a fase relativa contínua (FRC) média e contínua para os pares cotovelo-ombro e ombro-quadril. Os idosos têm FRC média e estabilidade reduzida em comparação com adultos mais jovens, e a velocidade afeta a FRC ao longo do tempo. A estabilidade da coordenação dos membros superiores foi maior para os idosos nas velocidades de caminhada preferidas e rápidas. 3) verificar a relação entre coordenação bilateral e coordenação intersegmentar em 24 idosos saudáveis (66,3±4,4 anos) e as relações entre os fatores. Medimos a coordenação da marcha através do índice de coordenação de fase (ICF), fase relativa contínua (FRC) e sua variabilidade. Os idosos têm ajustes na FRC, principalmente entre quadril e joelho, para melhorar a ICF. O controle intersegmentar desempenha um papel crucial na coordenação bilateral. Os resultados mostraram que os determinantes da coordenação parecem ser os principais determinantes da marcha do idoso por meio do equilíbrio corporal que é influenciado por diferentes aspectos. As habilidades físicas também podem determinar a marcha de idosos, apoiada pela velocidade da marcha, que aparece indiretamente como determinante da capacidade de condicionamento por meio do IRL.

Palavras-chave: Idosos; Marcha; Envelhecimento; Locomoção; Coordenação da marcha.

ABSTRACT

Older adults have impaired gait pattern, decreased speed, muscle weakness, impaired motor control, impaired limb coordination, decreased angles and stride length, increased double support time, and postural impairments. However, there is still no clear understanding of the effects of aging on the main determinants of gait. The overall aim of this Thesis was therefore to examine the functional and coordination determinants of gait in older adults. To achieve this goal, three original studies were developed: 1) explore the physical determinants of the decline in gait speed in 187 active older adults (72.2±6.8 years). Walking performance was characterized by 2 parameters: locomotor rehabilitation index (LRI) and walking ratio (WR). With advancing age, the self-selected and maximum speeds decreased by up to 26%, as well as the physical fitness variables worsened with aging. We found WR and body balance as predictors of maximum velocity, suggesting that it would be necessary to decrease WR to avoid imbalance. Excessive reductions in the analyzed parameters indicate loss of homeostasis in gait mechanics. 2) understand the extent to which upper limb control mechanisms are present during gait. 20 older adults and 13 young people walked at different speeds on a treadmill and mean and continuous relative phase stability (CRP) were calculated for the elbow-shoulder and shoulder-hip joint pairs. Older adults have reduced mean CRP and stability compared to younger adults, and speed affects CRP over time. The stability of upper limb coordination was greater for the older adults at preferred and fast walking speeds. 3) verify the relationship between bilateral coordination and intersegmental coordination in 24 healthy older adults (66.3±4.4 years) and the relationships between the factors. We measured gait coordination through the phase coordination index (PCI), continuous relative phase (CRP) and their variability. Older adults have adjustments in CRP, especially between hip and knee, to improve PCI. Intersegmental control plays a crucial role in bilateral coordination. The results showed that the coordination determinants seem to be the main determinants of the older adults' gait through body balance that is influenced by different aspects. Physical abilities can also determine the gait of older adults, supported by gait speed, which appears indirectly as a determinant of conditioning ability through the LRI.

Keywords: Older adults; Gait; Aging; Locomotion; Coordination of gait.

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SUMMARY

CHAPTER 1

1.1 GENERAL PRESENTATION

1.1.1 Contextualization and delimitation of study

This work represents the union association of two research groups that indirectly seek a common line of research. The first line of research is related to the research group LOCOMOTION - Mechanics and Energetics of Terrestrial Locomotion (Mecânica e Energética da Locomoção Terrestre / UFRGS, Brasil) under the coordination of my advisor, Prof. Leonardo Alexandre Peyré-Tartaruga. The group 's main objectives are to study the basic mechanisms of human locomotion under different conditions, environments and populations. The second line of research is related to the research group CELARI (Centro de Estudos de Lazer e Atividade Física do Idoso) coordinated by my co-advisor Prof. Andrea Kruger Gonçalves, seeking to investigate the physical and functional aspects related to aging, health and physical activity.

The study of the older adults population is strictly connected to my academic experience. The interaction with the different lines of research took place since undergraduation, where I sought to respond to the concerns that arose with professional practice in the extension/outreach project with the older adults, generating different results, including this thesis. The concern that gave rise to this work emerged in mid-2016 with the concern of the increase in the average age of participants in the extension project and the difficulties and changes in gait already perceived in everyday life. Perhaps it may be a speculation, but the more you observe the characteristics of the gait, the more you notice the intra-subject differences, it seems to be such a researcher's gaze. Thus, an experimental study was designed with different physical exercise interventions, seeking to understand the specific benefits of each intervention on the older adults 's gait. And, it was done, at the end of 2018 the project was qualified and the procedures for the execution of the project began. After all necessary steps, the study would start in April 2020. I don't need to say what happened...the pandemic was launched, and all activities were suspended. Thus, several meetings and attempts to organize the study were initiated, until it was necessary to decide and define viable alternatives for the period of time we were living. With the commitment, patience and help of both groups, it was possible to arrive at this representative study that investigates the different aspects of gait in the older adults, from questions of gait speed to aspects of motor control. A *zoom-out* of the changes brought about by the aging process.

One way to illustrate the purpose of this study is the number of people involved. In total, information from 244 older adults and young who came from the university's extension and research groups was analyzed. Additionally, approximately 50 researchers from different levels of training who somehow contributed to these studies. Thus, we have unique and singular characteristics, being a growing and innovative study in locomotion and aging.

1.1.2 Thesis structure

This study is divided into six chapters.

The first chapter is introductory and presentational to the study. The chapter presents a general introduction to the study, which addresses issues related to population aging, physical abilities, and their changes with aging, as well as aspects of physiomechanics that change with aging, and goals and hypotheses.

The second chapter presents a cross-sectional study. The first article explores how aging can be associated with a decline in spontaneous walking speed and seeks the main determinants of the speed decline across aging.

The third chapter presents a study of short communication. In this article, the objective was to compare the intersegmental coordination and the coordination stability of the upper limb between older adults and young people at different speeds (comparing also these speeds).

The fourth chapter presents a mechanistic study. This article studied the relationship between local and global coordination. In particular, we tested the relationship between bilateral consistency and coordination stability, and between bilateral accuracy and intersegmental coordination.

The fifth chapter summarizes the results of the three studies on the integrative point of view and the general conclusion of the thesis. The sixth chapter lists the abstracts and articles published during the doctoral period.

1.2 GENERAL INTRODUCTION AND RATIONALE OF THE THESIS

1.2.1 Aging process in the current scenario

The proportion of people over 60 years old is growing rapidly in the world, unlike other population groups. A 223% increase in the number of older adults is expected from 1970 to 2025. The life expectancy of newborns in 2050 is estimated at 83 years (UNFPA, 2012). The increase in this population is an expected phenomenon, resulting from demographic changes that occurred in previous decades, it is a process associated with changes in the epidemiological profile and in the social and economic characteristics of the population (CHAIMOWICZ, 2009). In Brazil, the older adults population grew rapidly, radically and ascending, when compared to the reduction in fertility levels and the increase in life expectancy at birth (WHO, 2005). The number of Brazilians over 60 years old in 2012 was 11%, statistical projections show that this number will be 19% of the population in 2030 (IBGE, 2011; BRASIL, 2018).

The aging process is an inexorable, dynamic and irreversible natural process, characterized by a greater vulnerability to aggression from the internal and external environment. It causes greater fragility in the body, leading to gradual and inevitable changes related to aging, regardless of whether the subject enjoys good health and a healthy lifestyle. The maintenance of the individual's autonomy and independence is crucial during aging (CIOSAK et al., 2011; FAN et al., 2016). Some theories explain this process, one of the best known is chronological and biological aging. Chronological aging will indicate the time elapsed since birth and is equivalent to the biological age for individuals who age normally. Biological aging is related to the decline of organic functions (e.g. reduced cell regeneration capacity and consequent tissue aging) (FERRUCCI et al., 2019).

The cycle of human organic functions reaches its maximum around 30-40 years, with stabilization at 40-50 years. After this period, a progressive functional decline of 1% per year begins (GRESPAN; GRAVINA, 2016). In addition to the influence of variables such as sex, origin, nutrition and life experiences, a link between the individual's adaptability and environmental aggression is necessary

(CIOSAK et al., 2011, VALER et al., 2015). Therefore, we must consider all the functional aspects of the aging individual, from physical and mental health to socioeconomic conditions and self-care capacity (UFMA, 2014; PAGAC, 2018).

Biological decline is a negative factor associated with the aging process, often accompanied by functional constraints. One of the major changes basically caused by biological losses is the decrease in functional capacity (MITCHELL, et al., 2012; PAGAC, 2018). Functional capacity is the ability to efficiently perform the activities of daily living that people should perform to take care of themselves and live independently and autonomously and a combination of the individual, their environments and the interaction between them (TUNA et al., 2009, WHO, 2015; FREITAS; COSTA; GALERA, 2016). In summary, there are biological changes in the musculoskeletal system (decrease in muscle mass and bone density), loss of agility, motor coordination, balance and joint mobility, and greater rigidity in cartilage, tendons and ligaments (RIKLI; JONES, 2013; STATHOKOSTAS et al., 2013). The decline in abilities and effectiveness in performing basic activities, such as walking, is accelerated when aging is associated with a sedentary lifestyle, directly affecting the individual's independence (MCKINNON; MONTEIRO-ODASSO; DOHERTY, 2015; BERG *et al.*, 2018).

Cardiorespiratory endurance decreases with aging and can be considered a risk factor for mortality (WILSON; TANAKA, 2000, PANTOJA et al., 2016). There is also a reduction in maximal heart rate and a decrease in cardiac functional capacity when subjected to exertion (DE VITTA, 2000). However, active older adults have better aerobic capacity than their inactive peers or sedentary young people, showing that exercise seems to delay some physiological processes that are affected by aging (HAYFLICK, 1997; PANTOJA et al., 2016).

Decrease in muscle mass (sarcopenia), muscle strength and power (dynapenia) are clear adaptations from aging (SHEPHARD, 2003, CLARK; MANINI, 2008; SIMS et al., 2013). Muscle strength peaks between 20-30 years, after 50 years there is a slight decline. From the sixth decade of life onwards, the loss intensifies and reaches 12-15% compared to its peak and 30-40% at 80 years of age (GURALNIK et al., 1995; METTER et al., 1997). Modifications in the musculoskeletal system affect muscle fibers, decreasing length and elasticity, tendons and ligaments lose elasticity and reduces the viscosity of synovial fluids. Muscle tissue suffers important losses, perceived with the reduction in growth hormone levels, contributing 40% of muscle tissue losses (NARICI; MAFFULLI, 2010).

Aging reduces conditioning capacities and coordinating functions such as body balance. Balance is regulated by three fundamental systems: visual, somatosensory and vestibular (ZEIGELBOIM et al.; 2004; TINETTI; WILLIAMS; GILL, 2000). It is a process that involves the reception/integration of sensory stimuli and planning and executing the movement for a stable posture and control of the center of gravity. For this, neuromuscular responses are needed to each new posture adopted, that is, a neural process that involves an organization of stability and an orientation of the body in space (WINTER, 1995; TIEDEMANN; SHERRINGTON; LORD, 2013). It is necessary to have a relationship between the internal and external forces that act on the body (DICHARRY, 2010). Factors that interfere negatively are the loss of muscle mass and changes in neuromuscular control (ACSM, 1998; ARAÚJO et al., 2013). Its main affections are related to the decrease in the ability to maintain the stability of the center of pressure, decrease in the individual's anticipatory and compensatory postural adjustments, which increases instability and greater sway to maintain the standing posture (PIZZIGALI, 2016; TIEDEMANN; SHERRINGTON; LORD, 2013; GONÇALVES et al., 2017).

1.2.2 Gait, mobility and coordination

The impairment in the gait pattern of the older adults is a subject that needs further studies because mobility has a crucial impact on quality of life. The most important outcome of the gait, which affects 20% of the older adults, is the reduced gait speed. Older adults people walk more slowly than young adults, which seems to be a compensatory strategy for stability, muscle weakness, impairment of motor control (BOHANNON, 1997; STOLZE et al., 2000; ASHTOM-MILLER, 2005; KO; HAUSDORFF; FERRUCCI, 2010), coordination disorders between the upper and lower limbs, reduced range of hip and ankle motion and postural difficulties (CRUZ-JIMENEZ, 2017; KEMOUNG et al., 2000). The step length and frequency are reduced, the time of double support increased, which constitutes a more stable and slower gait (SMITH; WEISS; LEHMKUHL, 1997; MIDDLETON; FRITZ; LUSARDI, 2015; HERSSENS et al. al., 2018).

The self-selected walking speed (SSWS) is a critical behavior attribute. While for people without motor restrictions, the SSWS is close to the most economical speed (lowest energy cost), older adults have SSWS lower than the most economical speed (SAIBENE; MINETTI, 2003). A reduced speed provides other changes that occur with age, decreased sensory acuity, changes in balance, decreased strength and reaction time (FAN et al., 2016). There is still no understanding that the speed decline is the result of a compensatory effort to improve safety or a consequence of the impaired muscle function, which may be linked to the loss of functional capacity *per se* (KO; HAUSDORFF; FERRUCCI, 2010). The gait speed is also representative of the survival and mortality of the older adults, as can be seen in the figures by Studenski et al. (2011), showing that the remaining years of life can be predicted from the gait speed, that is, the expected life expectancy for each sex and age increased as the gait speed was higher.

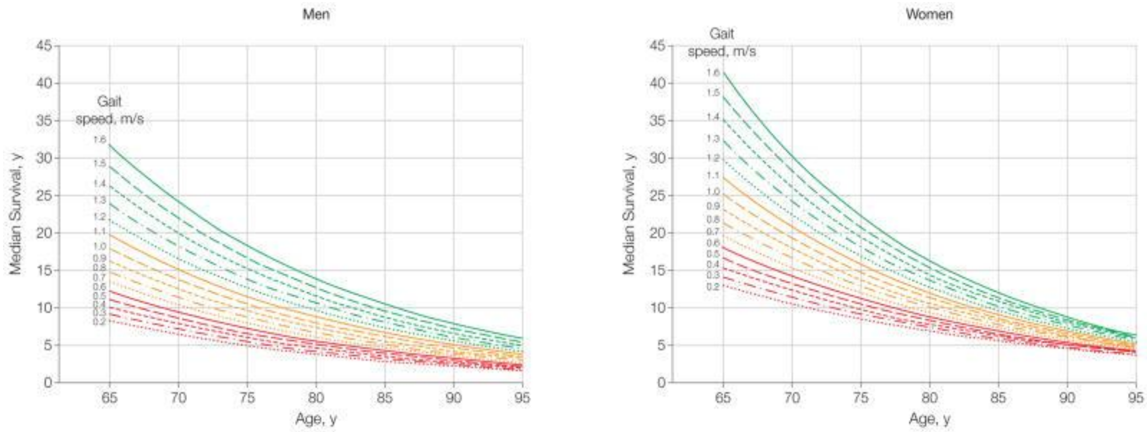

Figure 1.2.2.1 - Expected life expectancy per walking speed. Source: Studenski et al. (2011).

One of the most used definitions for the gait pattern of the older adults comes from the 60s, when scholars at the time showed interest in investigating the difference between young and old. The gait of the older adults was defined as a natural change, caused by the aging process and with the following characteristics: loss of speed, forward posture, rigidity when walking, shortening of the stride, larger base of support, en bloc trunk motion, and longer double support time (MURRAY et al., 1969; BARAK; WAGENAAR; HOLT, 2006; ABREU; CALDAS, 2008). The factors that contribute to this gait pattern are related to inadequate proprioception, slowness for the required postural responses, reduced range of motion and weakness in some muscle groups, especially the pelvic region and hip extensors (MURRAY et al., 1969; WINTER, 1995; KERRIGAN et al., 2003; KO; HAUSDORFF; FERRUCCI, 2010; CALLISAYA et al., 2011, FAN et al., 2016).

For healthy individuals, the preferred gait will be one that minimizes the energy cost of walking. The combination of impaired mobility with lower performance affects different characteristics of walking in the older adults. Among them is the consumption of metabolic energy and the cost of generating mechanical work (and the respective mechanical efficiency) can be determinants of functional mobility (CAVAGNA; KANEKO, 1977; ORTEGA; FARLEY, 2015), being affected by the hip extension angle, step width and frequency (WERT, et al. 2010). The total mechanical work is the sum of the work performed to elevate and accelerate the center of body mass in relation to the environment (external work), and the work performed to accelerate the body segments in relation to the body center of mass (internal work) (CAVAGNA; KANEKO, 1977). Compared to young people, the older adults have a greater internal work, although the total mechanical work is similar (MIAN et al., 2006). Differences in internal work seem to be attributed to anthropometric characteristics in body segments and the distribution of muscle mass (SCHUCH et al., 2011). Gait disturbances affect 35% of septuagenarians (CRUZ-JIMENEZ, 2017), reducing speed by approximately 0.2 ms^{-1} (HIMANN et al., 1988), which represents a decrease of approximately 15% per decade (GIMMON et al., 2017). In this age group, the lower metabolic economy and mechanical efficiency of walking are related to greater coactivation of antagonist thigh muscles during walking compared to young adults, which may be a compensatory action to the increase in joint stiffness and stability (MIAN et al., 2006; ORTEGA; FARLEY, 2015; MALATESTA et al., 2003). However, in octogenarians and nonagenarians, there are more intense alterations in locomotor patterns. For example, hip extension in the final support phase is especially impaired and seems to be related to the metabolic cost of walking (VANSWEARINGEN; STUDENSKI, 2014).

The balance capacity during gait involves maintaining the center of gravity and when it is projected beyond the limits of this base, which represents a challenge to the central nervous system. Among the losses related to gait balance in the older adults, the height of the foot in relation to the ground during the swing phase and the increase in frequency may be related to the weakness of the ankle and hip stabilizer muscles (LITTBRAND et al., 2009; LUGADE; LIN; CHOU, 2011). Maintaining balance seems to be essential for the quick response to external stimuli or disturbances, which can cause falls, and it is necessary to generate power in the ankle plantar flexors for balance recovery (FUJIWARA et al., 2011).

The gait considered normal is the result of a harmonious relationship between nervous, muscular and skeletal functions, any interference in some of these aspects will cause alterations in the individual's gait pattern (WINTER, 1995). Coordination is a process where degrees of freedom are organized in time and in a certain sequence for the reproduction of a functional movement pattern (BERSTEIN, 1967). Coordination is a class of measures of body movement composed of the time of one body segment in relation to another, for example the left-right pattern during a gait cycle. Young people have a bilaterally coordinated and symmetrical gait, changes in gait characteristics and coordination disorders with aging are risk factors for impaired mobility (JAMES et al., 2016; GIMMON et al., 2017). Impaired gait coordination was associated with worse mobility performance, regardless of the spatiotemporal gait variables (JAMES et al., 2016). Bilateral gait coordination was modified with changes in speed and showed a worse performance with older age (GIMMON et al., 2017).

Gait is an extremely complex motor skill composed of a sequence of cyclical movements of the lower limbs that provide opportunities for body displacement, added to a displacement of body mass and changes in speed and acceleration. This combination is the result of a harmony between external forces and internal forces from muscles, tendons, ligaments and joints (JUDGE; DAVIS; OUNPUU, 1995; GANVIR et al., 2012). The act of walking will require muscle strength and motor control, essential in a consistent functional action, to coordinate sensory input and muscle contraction (BYRNE et al., 2002, BISI; STAGNI, 2016). Looking specifically at movement coordination there are differences between young and old. Older adults present a less complex and potentially less flexible gait in response to disturbances, decline in mobility, muscle function and musculoskeletal health (VAN EMMERICK, 2005, PLOTNIK; HAUSDORFF, 2007, HAFER; BOYER 2018, DEWOLF et al., 2019).

Van Emmerick, et al., (2005) observed that all lower limb angles decreased with age, older adults had lower rotations of joint combinations (such as hip-shoulder) and at fast speeds coordination was out of phase. In addition to lower variability, greater asymmetry, and lower arm swing amplitude (MIRELMAN et al., 2015; DEWOLF et al., 2019). Some changes in gait characteristics, such as variability in spatiotemporal properties, can identify mobility and peripheral neuromuscular impairments (reduced ankle strength and range of motion) (JAMES et al., 2016). These changes affect intra-limb (between segments of a limb) and inter-limb (between right and left limbs) coordination (HADDAD et al., 2006, LI et al., 2005; FIELD-FOTE & TEPAVAC, 2002). Knowing the changes caused in coordination, the older adults should seek better coordination to allow compensation for deficiencies in strength, speed and range of motion (JAMES et al., 2016).

A common indication in the older adults' walk is increased variability, which is associated with body instability and, consequently, an increased risk of falls. Variability, defined as the fluctuation in gait characteristics from one step to another, is a natural and important characteristic of movement, which can characterize a rigid and immutable system (low variability) or an unstable and noisy system (high variability). Extreme situations can characterize systems that are less adaptable to disturbances, associated with degenerative processes or diseases (STERGIOU; HARBOURNE; CAVANAUGH, 2006, HERSSENS et al., 2018). Winter (1984) demonstrated that the intrapersonal variability of patterns of coordination of joint moments throughout the step period tends to be high at the knee and hip, but low at the ankle. This can influence the control of walking, as the variability of the step width can show differences between young and older adults during locomotion on the treadmill (OWINGS; GRABINER, 2003). Brach et al. (2007) confirmed the hypothesis that central nervous system deficiencies are manifested in increased support time variability at slow walking and sensory deficiencies in increased step width variability at higher speeds.

A literature review (HERSSENS et al., 2018) investigated the differences in spatiotemporal and gait variability measures throughout healthy adult life, highlighting that with aging, changes in the musculoskeletal system, peripheral and central nervous system arise. in dysfunctions in mobility and gait performance. In summary, the results suggest that most spatiotemporal parameters differ between age groups, older adults show a reduction in preferred walking speed, frequency, step and stride length, all related to more cautious gait. The authors emphasize that with increasing age, the older adults adopt compensatory strategies to supply an increase in postural stability, a weakening of the hip extensors and ankle plantar flexors, balance problems and a decline in central motor control.

1.2.3 Clinical indicators for gait assessment

The description or estimation of the spatiotemporal parameters of gait requires accurate models that are often not available outside the laboratory environment. The use of indices presents simple but relatively accurate models, being an effective and accurate way to track walking patterns and assess their changes.

1.2.3.1 Locomotor Rehabilitation Index

The Locomotor Rehabilitation Index (LRI) is a simple and integrative assessment method to assess gait that takes into account the effect of size across leg length. Numerically, it assesses how close the self-selected walking speed (SSWS) is to the optimal walking speed (OWS). OWS occurs when the gravitational potential and kinetic energy exchanges are optimized during each step (PEYRÉ-TARTARUGA; MONTEIRO 2016; GOMEÑUKA et al., 2019). Individuals may have a SSWS lower than the OWS, in this case, the metabolic cost of walking is typically increased (SAIBENE; MINETTI, 2003). The closer the SSWS is to the OWS, the more economical the walk becomes (PEYRÉ-TARTARUGA; MONTEIRO, 2016). However, high LRI values may be accompanied by lower OWS values than SWSS and the same can happen inversely (BARBOZA; WEIZEMANN; CARVALHO, 2021). The image (figure 1.2.2.2) presented by Peyré-Tartaruga; Coertjens (2018), simulates this behavior of the LRI after an intervention in a patient with locomotor impairment, the authors suggest that the increase in the LRI is due to improvements in SWSS, representing an increase in the patient's tolerance to sustain higher intensities in more economical speeds.

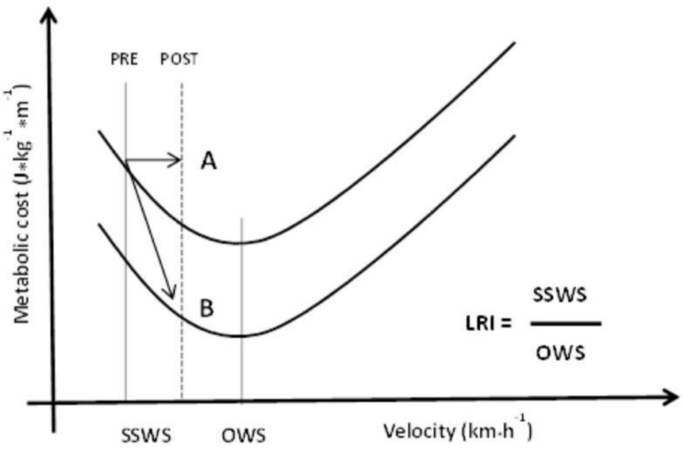

Figure 1.2.2.2 - Metabolic cost at different speeds and two possible training effects for a person with a locomotor disability. Source: Peyré-Tartaruga; Coertjens, (2018).

The index can be used to define the locomotor capacity, taking into account the fundamental concepts of terrestrial locomotion. The LRI has been used for different populations, patients with heart failure (FIGUEIREDO et al., 2013), Parkinson's (MONTEIRO et al., 2017; NARDELLO; BOMBIERI; MONTE, 2020) and the older adults (GOMEÑUKA et al., 2019). For heart failure patients, the improvement of the LRI provided greater savings in locomotion (FIGUEIREDO et al., 2013). While the LRI of Parkinson's patients was lower when performing a dual task in different directions (NARDELLO; BOMBIERI; MONTE, 2020) and more sensitive to the modifications provided by training with poles (MONTEIRO et al., 2017). This change was also observed in the older adults, being related to the increase in SWSS in response to training in Nordic walking and a longer stride length, which provides an optimization of energy exchanges in the center of mass, reducing the total mechanical work (GOMEÑUKA et al., 2019). In these situations, the lower the LRI values, the greater the rehabilitation potential, in which, indirectly, the integrative index shows that the individual walks with a substantial metabolic cost due to the deteriorated pendular mechanism (PEYRÉ-TARTARUGA; MONTEIRO, 2016). Thus, the use of LRI is encouraged as a new proposal to assess gait functionality, in addition to its applicability in locomotor rehabilitation, with information related to walking energy consumption (FIGUEIREDO et al., 2013; PEYRÉ-TARTARUGA; MONTEIRO, 2016; GOMEÑUKA et al., 2019).

1.2.3.2 Walk ratio

The walk ratio (WR) can be seen as a general index of gait kinematics, which simply describes the spatial and temporal coordination of gait (SEKIYA; NAGASAKI 1998; ALMARWANI, et al., 2019). WR is the ratio between the amplitude and frequency of rhythmic leg movement during walking, i.e., the ratio of step length and frequency (WR: cm / steps / min). Gait speed is measured by step length and step frequency (cadence), and its changes come from the variation of these gait characteristics (ROTA et al., 2011; BOGEN et al., 2018; FANG et al., 2019).

Figure 1.2.2.3 - Illustration of small steps at a high step frequency (left) or longer steps at a lower step frequency (right). The WR will represent the different combinations of stride length and stride frequency that can be selected to walk at a given speed. Source: Niay et al. (2020).

The walking speed, specifically, is determined by the product of the step length and the step frequency, therefore, it is possible to walk with infinite combinations of step length and frequency (SEKIYA; NAGASAKI, 1998). Fang et al. (2019) made a brief review and showed that the index is independent of walking speed in some studies, while it reduced with increasing speed in other studies. WR was stable for most speeds except very low, i.e. an invariable gait characteristic, allowing an ideal balance between energy expenditure and forward propulsion, for stability and for minimal attention demand during the walking (BOGEN et al., 2018). Almarwani et al. (2017) showed in studies from the 1960s that the WR did not vary between people who walked at various preferred speeds, the preferred combination of step length and frequency will coincide with the minimum energy expenditure and is optimal in terms of temporal and spatial variability and demand for attention for a given speed (ROTA et al., 2011). The optimal walking WR for adults is about 0.0064 (m / steps / min), but factors such as neurodegenerative disease and aging can induce a lower WR (SEKIYA; NAGASAKI, 1998; ALMARWANI et al., 2017). Low WR suggests a strategy to maintain speed, usually there is an increase in frequency while decreasing stride length; and high WR represent increased step length or lower frequency (BOGEN et al., 2018).

Given such importance, the WR has been useful and reliable for detecting early signs of gait deterioration, assessing gait patterns with aging, neuromotor gait control and gait coordination (ROTA et al., 2011; ALMARWANI et al., 2019; ALMARWANI et al., 2017), providing adequate health care (SEKIYA; NAGASAKI, 1998, KALRON et al., 2019).

1.2.3.3 Continuous relative phase

Joint coordination plays an essential role in maintaining a stable gait pattern. The segment coordination represents the patterns used to produce joint angles and these coordination patterns will provide information about the time and magnitude of the movements, representing the organization of various degrees of freedom in a simpler control strategy (HAFER; BOYER, 2018). The analysis of movement, through the theory of dynamical systems, has been widely used, as it provides a theoretical framework to simplify and work with complex systems (LAMB; STÖCKL, 2014). Dynamic systems can be composed of many interacting parts, as most movements involve a large number of moving parts that coordinate together and their behavior can be described by a single term or low-dimensional measure (LAMB; STÖCKL, 2014; GLAZIER; DAVIDS, 2009).

A continuous relative phase (CRP) is a continuous measure describing the space-phase relationship between two segments or joints throughout a gait cycle. Traditionally used to quantify the coordination and variability between two or more segments/joints during an activity (PETERS et al., 2003; CHIU; CHOI, 2012; LAMB; STÖCKL, 2014; HAFER; BOYER, 2018; IPPERSIEL; ROBBINS; DIXON, 2021). CRP represents the coupling of the movement phases, where it can be totally in phase (more strongly coordinated) or out of phase (less strongly coordinated) (HAMIL et al., 1999; IPPERSIEL; ROBBINS; DIXON 2021). The graph (figure 1.3) represents the time course of the knee-hip CRP (IPPERSIEL; ROBBINS; DIXON 2021).

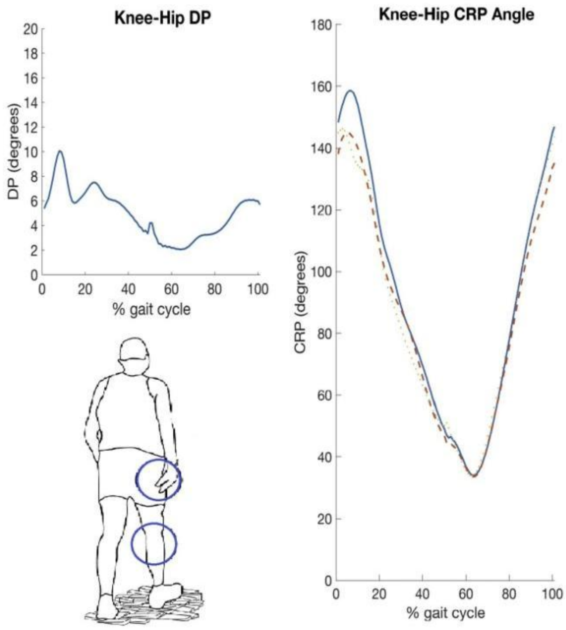

Figure 1.2.2.4 - Representative steps of the continuous relative phase (CRP). Knee-hip CRP angles on a 0-180 degree scale and Knee-Hip Deviation (DP) Phase, which quantifies CRP variability. Source: Ippersiel; Robbins; Dixon (2021).

The use of movement coordination measures and their variability, which show the organization of gait patterns, can reveal which mechanisms are affected by the natural physiological alterations of aging and how they cause alterations in gait mechanics. From this, it is possible to determine to what extent movement coordination differs with age and where these differences are located (HAFER; BOYER, 2018). Specifically, interjoint coordination patterns and their variability will provide essential elements for the understanding of neuromuscular control during functional movement. Thus, a reduction in the variability of segment coordination, due to lower coordination patterns, increases the risk of falls in the older adults, due to a smaller variety of movement patterns that generates a limitation of solutions for obstacles or stumbling blocks (CHIU; CHOU, 2013; IPPERSIEL; ROBBINS; DIXON 2021). Excessive variability of the supporting limb and reduced variability of the swinging limb in the knee-ankle interjoint coordination of older adults fallers may contribute to the risk of imbalance or tripping during walking (CHIU; CHOU, 2013).

CRP can assess a higher order of neuromuscular control which makes the continuous relative phase an attractive and popular variable for inter- and intra-limb coordination and can provide information behind age-related gait changes (CHIU;

CHOI 2012; LAMB; STÖCKL, 2014; HAFER; BOYER, 2018; IPPERSIEL; ROBBINS; DIXON 2021).

1.2.3.3 Phase Coordination Index

Healthy human gait is characterized by an anti-phasic left-right step pattern. The performance of movements involves gait coordination, the ability to properly time the step patterns from left to right within a stride, between the limbs and is fundamental for many tasks of daily living (PLOTNIK et al., 2013; PLOTNIK; GILADI; HAUSDORFF, 2007; SWANSON; FLING, 2018; JAMES et al., 2017).

The Phase Coordination Index (PCI) is a relatively recent and unique metric to quantify lower limb coordination (SWANSON; FLING, 2018). PCI combines two measures that represent the relative time of the contralateral heel strikes (figure 1.4), thus determining the phase, represented as phi (φ), that is, it combines the precision and consistency of phase generation in relation to the value of 180 ° (PLOTNIK; GILADI; HAUSDORFF, 2007; SWANSON; FLING, 2018). Thus, lower IPC values reflect a more consistent and accurate phase generation, being related to different health conditions, while higher values represent more impaired bilateral gait coordination (PLOTNIK et al., 2013; SWANSON; FLING, 2018).

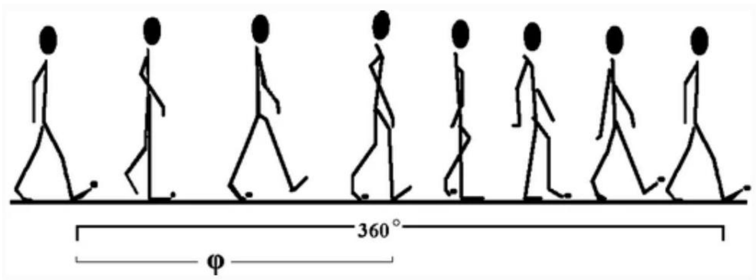

Figure 1.2.2.5 - Illustration determining the phase of the step of one leg in relation to the gait cycle determined by the other leg. The time difference between consecutive heel strikes of the two feet, normalized for the duration of the gait cycle, is defined as the phase. Source: Plotnik et al., (2007).

Bilateral gait coordination measures can distinguish important limitations involved with functional mobility. Increasing age has been associated with lack of coordination, causing deficits in the ability to coordinate two limbs to complete specific tasks, which requires the ability to control both legs in time and space

(PLOTNIK et al., 2013; JAMES et al., 2017; SWANSON; FLING, 2018). Older adults had increased PCI values compared to young people, suggesting that aging can limit the accuracy and constancy of the ability to time the gait cycle of one leg in relation to the other, which compromises the creation or execution of a symmetric gait (PLOTNIK; GILADI; HAUSDORFF, 2007).

1.3 JUSTIFICATION

Gait has been shown to be an important predictor of functionality in the older adults; it is a fundamental part of activities of daily living and an important marker of health, functional independence, quality of life and clinical evolution under different conditions. It is known that older adults with modified gait parameters may have greater difficulties in performing their daily activities.

Walking is an important component to assess and monitor the functional profile of individuals in different age groups, regardless of health status. Knowing the importance of walking as a marker of health and functionality status and that the spatiotemporal parameters, such as speed, frequency and stride length, adjust according to the physical condition of the older adults to ensure efficient movement, the clinical assessment of these parameters it becomes essential for the maintenance of functional capacity and the identification of groups of older adults with similar care needs and prognoses.

Functional capacity is totally linked to efficiency in the daily physical demands of the older adults, which ranges from the most basic activities to the most complex activities of the daily routine, and gait or gait performance is directly related to such demands. The assessment of the determinants of gait, or of the different parameters that go beyond walking, is essential to determine the parameters of health, frailty, and morbidity and mortality rates in the older adults. One of the examples of these parameters widespread in clinical evaluation is gait speed, as slow speeds are typical in older adults, and it has been recognized for a few decades as the sixth vital sign, being an indicator of functional capacity, general health, response rehabilitation, frailty, falls and countless other aspects related to the health of the older adults.

From this, in this study, the gait speed and the other parameters involved in gait will be discussed recurrently to seek answers to the proposed objectives. As well as helping to understand the effects of aging on gait, providing important knowledge for the clinic to identify the characteristics of a safe, reliable, independent, efficient and healthy gait. In a broader aspect, this thesis seeks to launch possibilities for future studies that seek to involve physiomechanics in an aspect closer to the clinic and interventions aimed at the older adults, seeking confirmation of the changes and characteristics of the population aging process.

1.4 OBJECTIVES

1.4.1 General objective

The objective of this thesis is to identify the functional and coordination determinants of gait in older adults.

1.4.2 Specific aims and hypothesis

The first specific objective is related to study 1, to investigate the motor skills that determine the speed and frequency of walking in older adult women. We hypothesize that the decrease in walking speed with aging is related to motor skills, that is, the locomotor rehabilitation index reflects the skills that represent power and the walking ratio represents those of control.

The second specific objective is related to study 2, to compare the continuous relative phase patterns in the upper extremity segment and its variability between older adults and young at different speeds. Also, determine whether there is a difference in continuous relative phase patterns in the upper-end segment and its variability at different velocities. Our hypothesis is that aging would reduce the continuous relative phase of the upper limbs and impair the stability of the older adult's coordination.

The third specific objective is related to study 3, to verify the relationship between bilateral coordination and intersegmental coordination in physically healthy older adults, during normal walking at different speeds, and to investigate possible relationships between coordination, bilateral consistency and precision, and coordination stability. Thus, we hypothesize that bilateral coordination can be explained by intersegmental coordination and that bilateral consistency is correlated with coordination stability, while bilateral precision is correlated with intersegmental coordination.

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CHAPTER 2

STUDY 1

Determinants of age-related decline in walking speed in older women

2.1 INTRODUCTION

Walking speed is frequently investigated in the older adult population. Aging, even under healthy conditions (MICHEL; SADANA, 2017), is associated with visible stiffness in ambulation, more cautious gait, and quantitative changes in virtually all walking parameters. Such changes include shorter step length and frequency (hence lower speed), larger step width, reduced trunk mobility, and higher risk of falls (MIAN et al., 2006; ABOUTORABI et al., 2016; HERSSENS et al., 2018; SCHOENE et al., 2019). Spontaneous walking speeds below 1.0 m s^{-1} are associated with increased mortality (CESARI et al., 2005; FIGGINS et al., 2021).

Declines in the most various bodily functions may contribute to alterations in walking performance. Cardiorespiratory endurance reaches its maximum capacity at about 20 years of age. Thereafter, until 65 years of age, there is a 20–30% reduction in cardiac output (ERKKOLA et al., 2021). Maximal muscle strength shows a reduction of 15–30% every 10 years after the fifth decade of life (PAPADOPOULOU, 2020; MANINI; CLAR, 2012), which is equivalent to a reduction of about 1% per year (PANTOJA et al., 2016). Muscle power production may also decrease as a result of mitochondrial dysfunction (see CONLEY et al., 2007). Changes in joint flexibility may explain the lower range of joint excursions, subtended mainly by loss of joint cartilage and a widespread decrease in collagen concentration, entailing loss of compliance and elasticity of joint capsules, ligaments, and tendons (KOTHARI et al., 2016; ERKKOLA et al., 2021).

Alterations in neural control also play a major role in age-related changes in walking mechanics (MIAN et al., 2006; ORTEGA; FARLEY, 2007). Decreased balance seems to induce changes in gait speed (CRUZ-JIMENEZ, 2017). A decrease in body balance may stem from the most various causes: delayed onset in muscle recruitment; atrophy of joint, tendon, and muscle receptors; loss of skills in postural responses; loss of proprioceptive fibers (SANDERS et al., 2019; GERARDS et al., 2021; MARTINA et al., 1998); and decreased stiffness of calf tendons, leading to delayed elongation of muscle spindles (ONAMBELE et al., 2006). The age-associated decline in static and dynamic balance variables related to postural sway (postural sway velocity, endpoint excursion, maximum excursion of the body's center of pressure) has been estimated at 1% per year (TAKESHIMA et al., 2014).

On the other hand, neural control can provide functional compensation for metabolic and dynamic losses, e.g., by over activation and co-activation of muscle patterns (ORTEGA; FARLEY, 2007; MILLER et al., 2021; DELABASTITA et al., 2021). A simple, more general form of adaptation is lowering the walking speed. This adaptation, however, is not without disadvantages. The muscular work per unit distance needed to translate the body system, represented by its center of mass, during walking (external work) is minimized by a passive transfer between kinetic and gravitational potential energy, like in an inverted pendulum (ALEXANDER, 2005). Maximizing the effectiveness of this mechanism, however, requires a given speed (CAVAGNA et al., 1976) and, for any given speed, a given step length and, therefore (speed being the product of step length and frequency), a given frequency (CAVAGNA; FRANZETTI, 1986). The optimal speed and (for that speed) the optimal step length is very close to those spontaneously adopted by adult humans, in the order of 1.4 m s^{-1} 0.65 m, respectively. Lower or higher speeds imply higher external work and metabolic expenditure per unit distance (cost of gait) (TESIO et al., 1991). For any given speed, increasing frequency implies a higher muscular work to reset, at each step, the limbs with respect to the body center of mass (internal work) (WILLEMS et al., 1995).

Studies assessing spontaneous walking speed in older adults have obtained contradictory results that seem highly sample-dependent (HERSSENS et al., 2018). Speed measures range from 0.79 (BOULIFARD et al., 2019) to 1.34 m s^{-1} (FUKUCHI et al., 2019). Naturally, speed needs to be normalized by body height, e.g., through the dimensionless Froude number (speed divided by the square root of height multiplied by gravity acceleration) (CAVAGNA; FRANZETTI, 1986). But this would simply lead to the conclusion that, in general, the spontaneous walking speed of older adults is lower than that of younger adults. The discrepancies across studies, however, suggest that metabolic efficiency is not the only constraint to walking speed.

Are there differences in this respect between older women and men? This is an intricate issue: walking habits (LEUNG et al., 2009) and body size may interact with age in determining both walking speed and frequency. Once these factors are taken into account, however, the overall picture seems one of a similar age-related decline in speed, with women adopting a lower step length (and hence a higher frequency) than men (FRIMENKO et al., 2015). Factors affecting step length, such as muscle strength and balance, are candidates to explain the reduced functional mobility of older women.

Clinical experience tells that, for sedentary individuals, walking to the point of muscular and cardiovascular fatigue, which would enhance metabolic efficiency, is not common. Why, therefore, do healthy older adults tend to adopt lower speeds even across short distances? It is still unclear whether motor capabilities are determinants of walking speed in older women. This study aims to investigate the motor capacities determining walking speed in older women.

2.2 MATERIALS AND METHODS

2.2.1 Study design

This is an open, cross-sectional study, carried out as part of a university outreach program in southern Brazil.

2.2.2 Participants

Untrained older women were recruited through community announcements and bulletins. Inclusion criteria for sample selection were age between 60 and 90 years, suitability for a complete assessment, community-dwelling status, and no regular physical training program in the previous three months. Exclusion criteria included the use of assistive mobility devices and any impairment limiting walking.

2.2.3 Assessments

Four tests were used to assess motor parameters that potentially influence walking mechanics. Tests were taken from the Senior Fitness Test battery (RIKLI; JONES, 1999) (see legend of Table 2.3.1 for short descriptions): (i) 8-foot up and go (agility/dynamic balance test, ABa), (ii) 30-second chair stand (lower body strength, LBS), (iii) 2-minute step (aerobic endurance, AE), and (iv) chair sit and reach (lower body flexibility, LBF). These tests have been extensively validated, do not require any special equipment, and can be easily applied in any clinical or exercise environment.

Self-selected (or, preferred) walking speed (SSWS) is related to body height or leg length (MALATESTA et al., 2003, LOWRY et al., 2012). For the measurement of SSWS, participants were asked to walk, in an indoor ambience, at their preferred usual speed between two points marked on the floor at 10 m distance, three times for each modality. The mean of three repetitions was used for further analysis (NOVAES et al., 2011). The same procedure was applied to determine the maximal walking speed (MWS).

The locomotor rehabilitation index (LRI) was also computed. This parameter is the ratio of the observed walking speed to the predicted optimal (lowest cost) walking speed (Peyré-TARTARUGA; MONTEIRO, 2016; GOMEÑUKA et al., 2019). To predict a subject's optimal walking speed, data were normalized using the dimensionless Froude number (Fr), as shown in Eq. (1):

$$\mathrm{Fr} = v/(g \times L)^{0.5} \qquad (1)$$

where v is the speed, g the gravity acceleration, and L the lower limb length (measured from the anterior-inferior iliac spine to the ground through the lateral malleolus) (VAUGHAN; O'MALLEY, 2015). The dimensionless optimal walking speed (OWS, Eq. 2) in humans corresponds to Fr = 0.25 (Eq. 1). In a given individual,

$$\mathrm{OWS} = 0.25\,(g \times L)^{0.5} \qquad (2)$$

It is thus easy to calculate the LRI for SSWS (Eq. 3):

$$\mathrm{LRI} = (\mathrm{SSWS}/\mathrm{OWS}) \times 100 \qquad (3)$$

The LRI has been applied to assess different populations, including patients with heart failure (FIGUEIREDO et al., 2013) and Parkinson's disease (MONTEIRO et al., 2017), and older adults trained in Nordic walking (GOMEÑUKA et al., 2019).

The walk ratio (WR) was calculated as the ratio of step length to frequency (SEKIYA; NAGASAKI, 1998; ROTA et al., 2011; BOGEN et al., 2018; KALRON et al., 2020), with step length expressed in mm and frequency in steps min^{-1}. The normal WR for adults and older adults up to 85 years of age is in the order of 5.5–6.5 mm $step^{-1}$ min^{-1}, across a wide range of walking speeds and body heights. Of note, this parameter is consistently lower by about 5% in women compared with men (BOGEN et al., 2018). The WR serves as a sensitive indicator of neural and cognitive gait impairments: it significantly decreases in multiple sclerosis (ROTA et al., 2011; KALRON et al., 2020) and Parkinson's disease (ZANARDI et al., 2021) as well as in healthy subjects under high attention demands (ALMARWANI et al., 2019).

We applied predictive models for either SSWS or MWS and either LRI or WR. Given that multicollinearity is expected across variables describing a subject's motor performance (mostly between MWS and SSWS but also between speed and the derived index LRI), we opted to use a decision-tree model rather than a conventional multiple regression model.

2.2.4 Statistical analysis

SSWS, MWS, LRI, and WR data were tested for normality of distribution based on skewness and kurtosis and then summarized as mean (standard deviation, SD) and median (interquartile range, IQR) or median (IQR) when appropriate. Significance was set at $p < 0.05$, and p-values were Bonferroni-adjusted for multiple comparisons. A predictive regression model was applied using a recursive partitioning algorithm, i.e., a classification and regression tree (CART) model.

CART is distribution-free. It transforms continuous levels into ordinal grades. The algorithm builds a decision tree based on binary splits on variables (either continuous, ordinal, or categorical). At each split, nodes are generated, and these nodes can be further split. The algorithm automatically detects interactions (i.e., the tree/node structure) between independent variables, providing the highest explanation of variance for the dependent variable (either categorical or continuous; here, continuous). The final result (terminal nodes) comprises a series of classes with the lowest possible within-class variance and the highest possible between-class variance. Unlike conventional linear regression modeling, in which the analyst must specify the expected interactions, CART itself discovers interactions, even high-order ones that are very difficult to hypothesize (BREIMAN et al., 1984). The algorithm is more sensitive to interactions than to main effects. The model's variance explanation is much less vulnerable to multicollinearity issues. Each split is performed on a single variable. If no further information is added by further splitting on a covariate, the latter is ignored. Software packages typically allow the analyst to control the procedure by imposing a minimum number of observations on each node or by setting stopping rules for tree branching (for a simple clinical example, see D'ALISA et al., 2006). A priori knowledge or requirements can thus complement the purely algebraic search for the maximum amount of variance explained. The stability of the predicted model can be inferred either by imposing the model splits (from the building sample) to an independent (validation) sample or by simulating several independent samples (boot-strapping) originating from the available sample. This is typically done through random extraction of subsamples and substitution of their values by random replication of observations coming from the remaining sample or the original total sample (resampling). In any case, the amount of variance explained for the validated tree unavoidably declines (shrinks) with respect to the variance explained for the original sample. It is left to the analyst to decide whether the model is satisfactorily stable or not (BREIMAN et al., 1984). There is no rule of thumb for accepting a given amount of variance explained. A reasonable empirical threshold for the validation tree is 30%, as suggested by the results for trees effectively predicting the length of stay, care costs, and functional outcomes of rehabilitation inpatients in the USA (STINEMAN, 1995).

In the present study, CART analysis is initiated from unsplit dependent variables (SSWS, MWS, LRI, WR) (root nodes). Each of these variables was split into nodes according to optimal cut-off points for the remaining variables to maximize the variance explained. Splitting continued until terminal nodes were defined, building the final classification model. The limitations imposed on each tree were as follows: maximum splitting levels, 10; splitting algorithm, least squares; minimum size node to split, 10; minimum rows allowed in a node, 5; tree pruning and validation method, cross-validation; number of cross-validation folds, 10; and tree pruning criterion, within one standard error of minimum cost complexity.

Descriptive statistics and regression modeling were done using IBM SPSS® version 21.0 (IBM Corporation, USA), and STATA® software (Stata Corp. LLC, USA, version 16.0). CART analysis was done through DTREG® software (DTREG, Brentwood, TN-USA, 2021).

2.2.5. Ethics

The study was conducted in accordance with the protocol approved by the Ethics Committee of Research involving human beings from Universidade Federal do Rio Grande do Sul (project number: 17243819.0.0000.5347 and clinical trials ID:NCT04348539). All participants who agreed to participate signed an informed consent form. For the Istituto Auxologico Italiano, this study fell within the RESET research program, Ricerca Corrente IRCSC, Italian Ministry f Health.

2.3 RESULTS

Table 2.3.1 provides descriptive statistics and a short definition of all variables assessed in this study.

Table 2.3.1 - Descriptive statistics for the variables in the study.

	Mean/Median (SD/IQR)	**Range**
Age (years)	72.22 (6.8)	60;88
Height (m)	1.56 (0.06)	1.39;1.73
BMI	28.37 (4.67)	19.85;49.07
LRI (%)	90.0 (13.83)	60.1;120.7
WR	0.56 (0.52;0.63)	0.35;1.02
LBS (no. full stands)	16 (13;19)	6;30
LBF (cm)*	-3.44 (10.70)	-29;25
ABa (seconds)*	5.1 (4.52;5.65)	3.37;8.85
AE	87.50 (15.89)	53;128
SSWS (m s^{-1})	1.30 (0.22)	0.77;1.87
MWS (m s^{-1})	1.74 (0.30)	0.94;2.74

Source: Author's own elaboration.

Note: BMI, body mass index (mass height-2); LRI, locomotor rehabilitation index; WR, walk ratio; LBS, lower body strength (number of full stands in 30 s with arms folded across chest); LBF, lower body flexibility [from sitting position at front of chair, with leg extended and hands reaching toward toes, number of cm (+or −) from extended fingers to tip of toe; negative values: cm missing to toes contact]; ABa, agility/dynamic balance test (number of seconds required to get up from seated position, walk 8 foot, turn, and return to seated position on chair); AE, aerobic endurance (number of full steps completed in 2 min, raising each knee to point midway between patella and iliac crest -score is number of times right knee reaches target); SSWS, self-selected walking speed; MWS, maximum walking speed; * the lower the value, the better the condition.

Age, SSWS, MWS, LRI, WR, and the four independent variables (ABa, LBS, AE, and LBF) were tested for normality based on skewness and kurtosis (Bonferroni-adjusted $p < 0.006$). Only ABa, LBS, and WR were significantly non normal (data not shown); thus, the assumption of linear regression was violated. For each of these variables, observations smaller than or greater than three SDs beyond the mean were trimmed (for linear regression only, not for further analyses). WR ratio remained non normal ($p < 0.005$ for skewness and kurtosis), as it had a uniform distribution. Linear regression was applied despite this limitation.

Table 2.3.2 revealed that all variables had a significant tendency of worsening with age (with confidence limits never including zero). The change was not significant for AE. In any case, the worsening was moderate. From 60 to 88 years of age, SSWS and MWS worsened (i.e., declined) by 22 and 26%, respectively. For the other variables, worsening ranged from 14 to 33%. For LBF, a percentage change would be misleading: finger–toe distance increased from 0 to 8 cm. The amount of variance explained by age was low for all variables, exceeding 10% for WR, only.

Table 2.3.2 - Linear regression modeling of dependent variable (Age) versus independent variables.

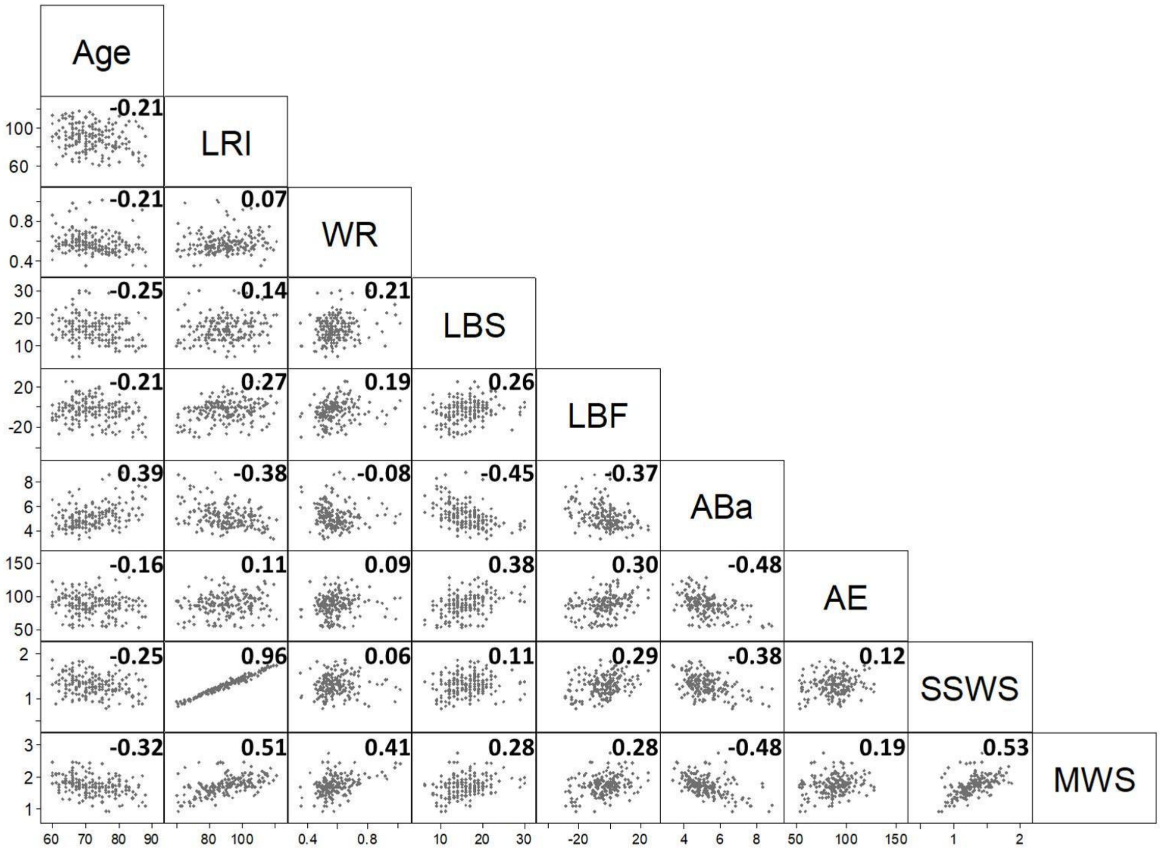
	n	β (95% CI)	const (95% CI)	R^2	$p^{\#}$	change§
LRI	185*	-0.419 (-0.711;-0.128)	120.2 (99.08;141.3)	0.04	**0.0051**	-14%
WR	187	-0.003 (-0.005;-0.001)	0.803 (0.641;0.966)	0.04	**0.0087**	-15%
LBS	185^	-0.158 (-0.245;-0.070)	27.30 (20.95;33.66)	0.06	**0.0005**	-33%
LBF	187	-0.331 (-0.555;-0.108)	20.49 (4.282;36.70)	0.04	**0.0039**	107%$^{\&}$
ABa	183^	0.048 (0.030;0.065)	1.718 (0.461;2.976)	0.14	**0.0000**	22%
AE	187	-0.364 (-0.700;-0.029)	113.8 (89.50;138.1)	0.02	0.0333	-12%
SSWS	187	-0.009 (-0.133;-0.004)	1.932 (1.600;2.266)	0.07	0.0002	-22%
MWS	187	-0.014 (-0.020;-0.008)	2.726 (2.283;3.168)	0.10	0.0000	-26%

Source: Author's own elaboration.

Note: LRI: Locomotor rehabilitation index, WR: walk ratio, LBS: lower body strength, LBF: lower body flexibility, ABa: agility/dynamic balance test, AE: aerobic endurance, SSWS: self-selected walking speed, MSW: maximal walking speed, β: slope coefficient of linear regression, const, y-intercept of linear regression; CI: confidence interval, R^2: proportion of variance explained; #: Bonferroni adjusted significance level 0.006, *: Two missing data for LRI, ^: observations exceeding the mean by ± 3SD were trimmed, §: Percent change from predicted values at 60 and 88 years, Positive changes indicate worsening for ABa and LBF, negative changes indicate worsening for LBS and AE, &: from 0 to -8cm.

The correlation matrix of the nine variables (Fig. 2.3.1) gives an overview of bivariate associations.

Figure 2.3.1 - The scatter plot provides the correlation half-matrix of the age, locomotor rehabilitation index (LRI), walk ratio (WR), lower body strength (LBS), lower body flexibility (LBF), agility/dynamic balance test (ABa), aerobic Endurance (AE), self-selected walking speed (SSWS), and maximal walking speed (MSW).

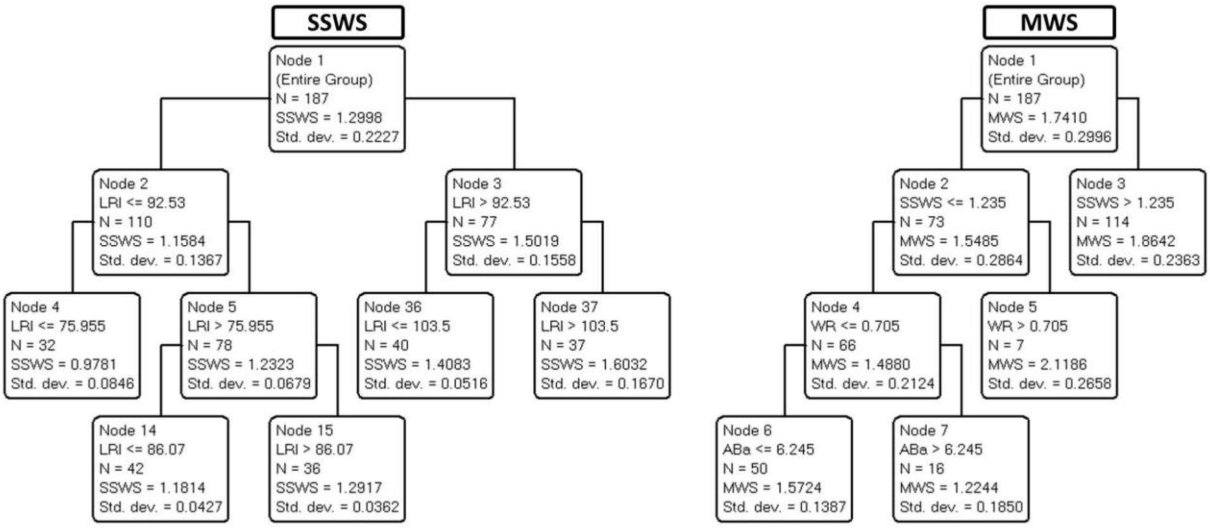


Note: Pearson's correlation coefficients are given in the corresponding boxes.

It must be highlighted that lower values of AE (cardiorespiratory fitness index) and LBF (joint flexibility index) indicate better performance. Fig. 2.3.1 shows that most of Pearson's correlation coefficients were very low. Only the correlation coefficients between LRI and SSWS (0.96), LRI and MWS (0.51), and SSWS and MWS (0.53) were higher than the arbitrary threshold of |0.5|. These findings were expected (see Eqs. 1–3), given that these variables are either derived from each other (LRI and MWS or SSWS) or strictly dependent on the subject's height (MWS and SSWS).

Interactions between multiple variables were explored through CART analysis. Fig. 2.3.2 depicts the decision trees used to predict SSWS (left panel) and MWS (right panel).

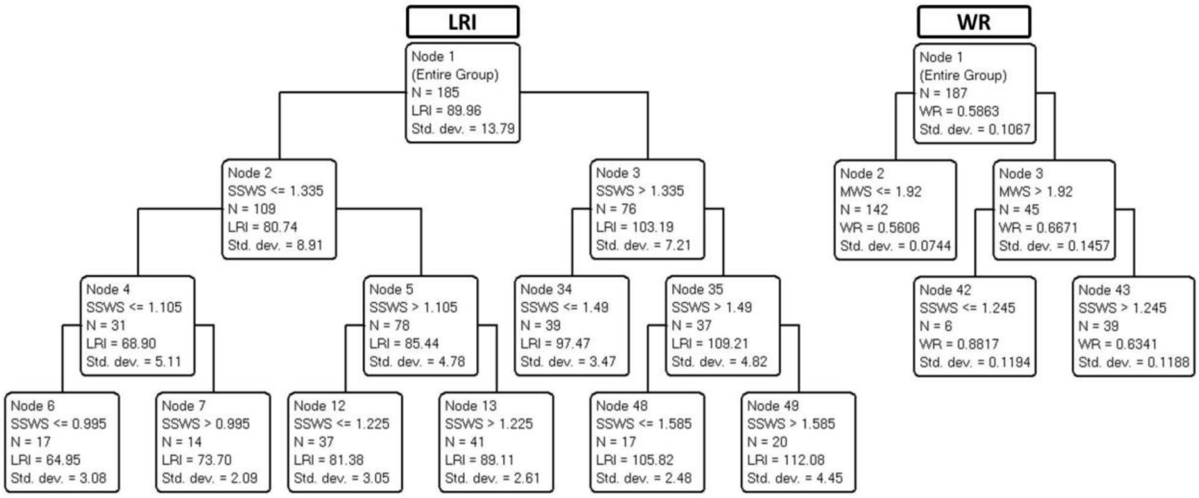
Figure 2.3.2 - Final classification and regression tree (CART) prediction model of self-selected (SSWS) and maximal (MWS) walking speed.



Note: For each "node" the figure gives the number of observations (N), and the mean and standard deviation (Std. dev.) of the corresponding sample, or split subsample. For each branch, the optimal cut-off score giving rise to splitting is reported. The main predictor of SSWS was the locomotor rehabilitation index (LRI). The main predictors of MWS were walk ratio (WR) and balance (ABa).

Fig. 2.3.3 shows the trees developed to predict LRI (left panel) and WR (right panel).

Figure 2.3.3 - Final classification and regression tree (CART) prediction models of locomotor rehabilitation index (LRI) and walk ratio (WR).



Note - Other information: see legend of Figure 2.3.2.

Table 2.3.3 summarizes the variance explained (for both training/building and validation data) for each of the four trees shown in Figs. 2 and 3.

Table 2.3.3 - Variance explanation of the decision trees for Figures 2.3.2 and 2.3.3.

TARGET VARIABLE	**Variance explanation %**	
	Training data	Validation data
Locomotor rehabilitation index	95%	93%
Walk ratio	33%	21%
Self-selected walking speed	84%	80%
Maximal walking speed	50%	36%

Source: Author's own elaboration.

As demonstrated by Table 2.3.3, the amount of variance explained by validation trees was satisfactory for SSWS, MWS, and LRI (ranging from 36 to 93%) but barely acceptable for WR (21%). The results suggest that most of the independent variables, including age, were not predictive of SSWS. In the corresponding tree, only LRI was retained, a circular finding (see above). By contrast, MWS was explained by SSWS (another expected finding) and, notably, by WR for speeds below 1.23 m s^{-1} as well as by ABa for WR values of less than or equal to 0.7 (most of the cases).

2.4 DISCUSSION

The expected associations between SWSS and MWS did not convey meaningful information. The association between SWSS and LRI, although expected, indicates that 7% of the variance in SWSS is related to size effects, and, therefore, LRI seems to be an improved marker of functional mobility due to size-dependent variation in gait speed. Other points deserve consideration. Neither age, nor any of the four motor indices selected (LBS, LBF, Aba, and AE, see legend of Table 2.3.1), nor WR explained SSWS. MWS was partially explained by the interaction between WR and ABa (Fig. 2.3.2). The WR tree (Fig. 2.3.3) confirms the relationship of WR with speed. Do these "negative" and "positive" findings conceal useful information?

2.4.1 An algebraic explanation

It must be said that explanation of variance requires some variance to be explained and a covariance. Along a 28-year gradient, the spontaneous and maximal velocities of older women undergo small changes, with a high interindividual variation. WR is largely invariant with speed and age. Not surprisingly, the weak relationship between speed and age (Table 2.3.1) is lost if a bivariate association is abandoned in favor of an interactive model (Figs. 2.3.2 and 2.3.3). Of course, a greater sample size might have allowed obtaining a more branched and explanatory decision tree. This algebraic interpretation, however, does not seem to be entirely satisfactory. An interpretation based on physiology, from outside the data, should be considered based on numerical assumptions.

2.4.2 Looking for an explanation in physiology

The results suggest that healthy aging, at least in this sample of women, implies a mild tendency for a decrease in SSWS, unexpectedly unrelated to the various physical performance parameters analyzed and the step length/frequency ratio (WR). The question then arises: given that these women were capable, at various ages, of increasing their speed (on average, SSWS was 1.30 m s^{-1}, whereas MWS was 1.74 m s^{-1}), why did they not retain the same SSWS at all ages? A second unanswered question is how could WR remain unrelated to both age and the various motor performance parameters? After all, step length and frequency should reflect lower limb joint power, mobility, and balance.

One reason may be that human walking has very wide margins of safety. In symmetric gaits, overall energy expenditure is minimal, given the refined pendulum-like exchange of mechanical energy of the center of mass. This characteristic makes humans the most efficient walkers in the animal realm (SOCKOL et al., 2007; HENN et al., 2012). Not only cardiorespiratory power but also the power required to drive muscles (mainly, the plantar flexors) remains much below the ceiling level (TESIO et al., 2017). Despite having a more overall flexed posture, lower limb joint excursions retain wide mobility margins. In case of focal strength deficits, compensation may occur between limbs and, within the same limb, between joints (TESIO et al., 1991; TESIO; ROTA, 2019). Once the speed needs to be decreased (see below for a tentative explanation), there seems to be no need for taking longer and more frequent steps than that already foreseen for the new speed. In case of need, however, a wide margin of safety remains for decreasing WR. In fact, at any given speed, a decrease in step length has a minimal influence on the effectiveness of the pendulum mechanism until a 50% decrease is reached (CAVAGNA; FRANZETTI, 1986; TESIO; ROTA, 2008).

Therefore, as a form of speculation, it can be hypothesized that cardiac–energetic or musculoskeletal constraints do not determine the age-related decline in speed. Rather, as suggested by several authors (see Introduction), balance control may represent a hidden, relevant determinant of the mild age-related decrease in SSWS.

2.4.3 The role of balance compared with that of other walking constraints

Over short distances, one can well afford a mildly higher metabolic cost unless this is prevented by severe cardiac or respiratory deficits. However, because of its pendulum-like mechanics, the body center of mass must be accelerated forward, upward (CAVAGNA et al., 1976), and laterally (TESIO; ROTA, 2019) at each step to overcome ground friction and gravity acceleration; the greater the ground friction, the longer the step, and the faster the movement, the shorter the step duration. These mechanical demands decrease by reducing walking speed. In particular, such a decrease in speed leaves more time for the amazingly fast U-turn from one side to the opposite at each step, as demonstrated by the analysis of the 3D trajectory of the body center of mass during a single stance (MALLOGGI et al., 2021).

Once the speed is conveniently lowered, therefore, a further decrease in step length (as evidenced by a lower WR) would unnecessarily entail a higher "internal" work per unit distance, i.e., the muscular work needed to reset the limbs at each step (WILLEMS et al., 1995). Not surprisingly, WR remained nearly invariant with age and SSWS in the present sample of women, confirming literature data on a wide range of velocities and adult ages (BOGEN et al., 2018; ROTA et al., 2011). This invariance, however, does not hold for the maximum speed attained (MWS). Consistently with its explanatory role, balance is known to decrease in healthy aging. In the present study, ABa was the variable that most depended on age (Table 2). It entered the prediction algorithm of MWS together with WR only. These results point toward a pivotal role of balance in determining the decline of speed in aging, at least at higher speeds. Physiological and clinical knowledge supports this view. It should be noted that WR is diminished whenever the balance is primarily affected (see Introduction). At any given speed, WR decreases when walking on slippery surfaces (CAPPELLINI et al., 2010), and, as a rule, in the case of neural impairments. Furthermore, the higher co-activation of lower limb muscles (MIAN et al., 2006; GOMEÑUKA et al., 2020) may help to understand the role of balance in gait speed reduction in older women.

2.4.4 Aging and walking, and what LRI and WR tell us

To sum up, in healthy aging, the decrease in speed (either self-selected or maximal) is modest. LRI and WR, which are related to each other, are virtually stable and seem to be unrelated to cardiorespiratory and musculoskeletal performance. This is no surprise, given the high effectiveness of human bipedalism. LRI and WR indices provide complementary information. They both seem to reflect a homeostatic control of gait, so that alterations might represent alarming early predictors of latent cardiorespiratory or joint power limitations (LRI) and/or latent balance deficit (LRI and WR). In particular, a decrease in WR may indicate balance deficits insufficiently compensated for by a reduction of speed. In support of this speculation, one should consider that human bipedalism is unique among bipedal vertebrates in many respects. For instance, the dominant role of plantar flexion as the main "engine" of gait is unique (USHERWOOD et al., 2012). Another unique feature of particular interest is the need for a very refined balance control on the frontal plane (MALLOGGI et al., 2020; CASSIDY et al., 2014). This need can represent a weakness in the case of balance deficits and much other neural impairment, leading to a reduction in speed and, in the most severe cases, further reduction of step length.

Three limitations of the present study cannot be overlooked. First, the results refer only to women. Second, only short distances were tested; speed and LRI and WR indices might have differed at longer distances. Third, the sample size did not validate the predictive model in an independent sample, representing a complementary, and perhaps a more robust, and form of validation than cross-validation.

We concluded that walking speed, muscle strength, flexibility, and balance are reduced in women aged 60 to 88 years. Whereas LRI seems to denote physical capabilities, WR represents a key coordinative aspect of functional mobility particularly related to balance in older women. The results suggest that both LRI and WR are helpful as a short screening battery for walking performance in aging and, potentially, in disability. These indices, however, only measure the presence of complex, tenacious, adaptive, and homeostatic mechanisms so that any alterations should entail a deeper, causal diagnostic inquiry.

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CHAPTER 3

STUDY 2

Age-related changes in upper limb intersegmental coordination at different walking speeds

3.1 INTRODUCTION

Aging causes changes in gait control. Older adults adapt walking speeds with different muscle control strategies (CHIU; CHOU, 2012; DEWOLF et al., 2019). In particular, gait speed affects the pattern and variability of intersegmental coordination, especially when walking at slower speeds (CHIU; CHOU, 2012). The continuous relative phase (CRP) and the coordination stability of lower limbs are modified in older adults compared to young, particularly at midstance and terminal swing (HAFER; BOYER, 2018). Specifically, older adults present an anti-phase lower limb coordination pattern, while young adults show in-phase coordination patterns during the terminal swing (HAFER; BOYER, 2018). Conversely Ippersiel et al., (2021), showed older adults were more in-phase during the swing phase. Furthermore, both studies show young adults with lower limb coordination stability (HAFER; BOYER, 2018; IPPERSIEL; ROBBINS; DIXON, 2021).

Although older adults walk with a reduced trunk (VAN EMMERICK et al., 2005) and arm (MIRELMAN et al., 2015) movement, less is known about the changes in upper limb intersegmental coordination and CRP stability due to aging at different speeds. Therefore, we aimed to compare the mean and stability of CRP in upper limbs between older and young adults at different speeds. We hypothesized that aging would reduce the upper limbs' CRP and would harm older adults' coordination stability.

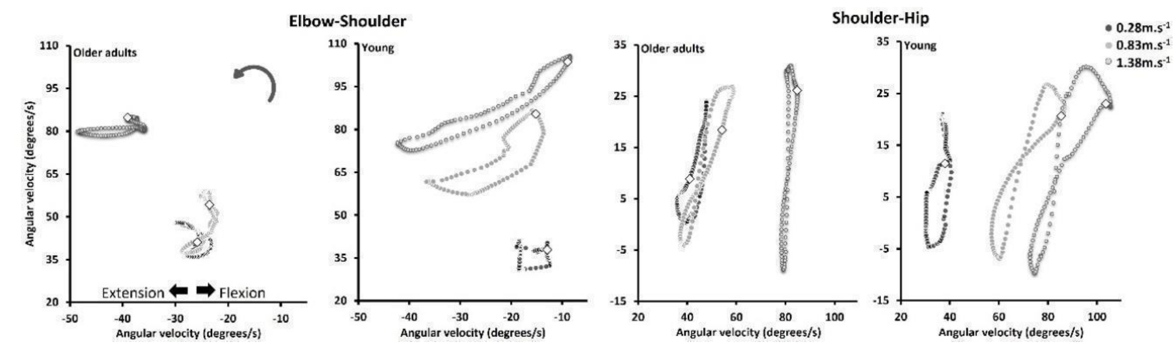
3.2 METHODS

This observational and cross-sectional study was approved by the local ethics committee (CAAE: 33784014.7.0000.5347). Participants read and signed an informed consent form. Thirteen young (07 men and 06 women, age: 29.5±4.7 y, mass: 75.5±9.6 kg, height: 1.72±6.24 m, means ± SD) and 20 older adults (07 men and 13 women, age: 66.4±4.3 y, mass: 77.2±14.2 kg, height: 1.65±9.20 m, means ± SD) were recruited from the community. Inclusion criteria included having good health, no known neurological or cardiopulmonary problems, no history of orthopedic, rheumatologic, or musculoskeletal disorders that prevented walking or assessment. The participants performed the walking tests on a motorized treadmill. Anthropometric data, CRP mean and coordination stability were assessed (for details, see Table 3.2.1).

We applied generalized estimating equation (GEE) models to examine the main and interaction effects (speed: 0.28, 0.83, 1.38 ms-1; and group: older and young adults), and Bonferroni post-hoc test (α = 0.05) in IBM SPSS® version 21.0.

Table 3.2.1 - Description of the procedures used.

Experimental protocol Three-dimensional kinematic data were collected by a motion analysis system using six infrared cameras. The three-dimensional reconstruction of the captured kinematic data was obtained automatically by the Vicon NEXUS® - *Vicon Motion Capture System (Oxford Instrument Group-USA, 1984)* Thirty-five reflective spherical markers were placed at anatomical landmarks of interest, and subjects wore dark sportswear and shoes suitable for physical activity - *Plug-in-Gait Full-Body model (NEXUS® 1.8.5)* **Walking test at different speeds** The familiarization with the treadmill followed the standard laboratory protocol. The different submaximal walking speeds were randomized, the subjects walked on a treadmill for 5 minutes at each speed (0.28, 0.83, 1.38 $m.s^{-1}$) with 5 minutes of rest, the kinematic collection was performed at the last minute - details in *Gomeñuka et al., [7].*
Intersegmental coordination and coordination stability Intersegmental coordination was evaluated from the continuous relative phase (CRP) and coordination stability from the CRP variability. It is noteworthy that coordination stability represents the inverse of variability. The CRP values were calculated using the Hilbert transform due to robustness of the method to frequency artifacts [8]. We determined the pairs of joints in the sagittal plane elbow-shoulder, shoulder-hip divided into 5 subphases of stance phase: - Total stance: CRP angle at all contact - Loading response: angle at contact 0-20% - Mid-stance: angle at contact 20-50% - Terminal stance: angle at contact 50-80% - Push-off: angle at contact 80-100% The processing data was performed with a custom routine implemented in LabView vs. 2017 (National Instruments, Austin, USA). The figure represents the phase portraits of the joint combinations.



Source: Author's own elaboration.

3.3 RESULTS

Differences in CRP of upper limbs between older and young adults for elbow-shoulder occurred during total stance, mid-stance and terminal stance; for shoulder-hip in loading response, mid-stance and terminal stance (Table 3.3.2). These differences were also observed in the coordination stability, except for loading response (Table 3.3.2).

Table 3.3.2 – Mean, standard deviation and statistical significance of for older adults and young during each functional period of the stance phase at different speeds.

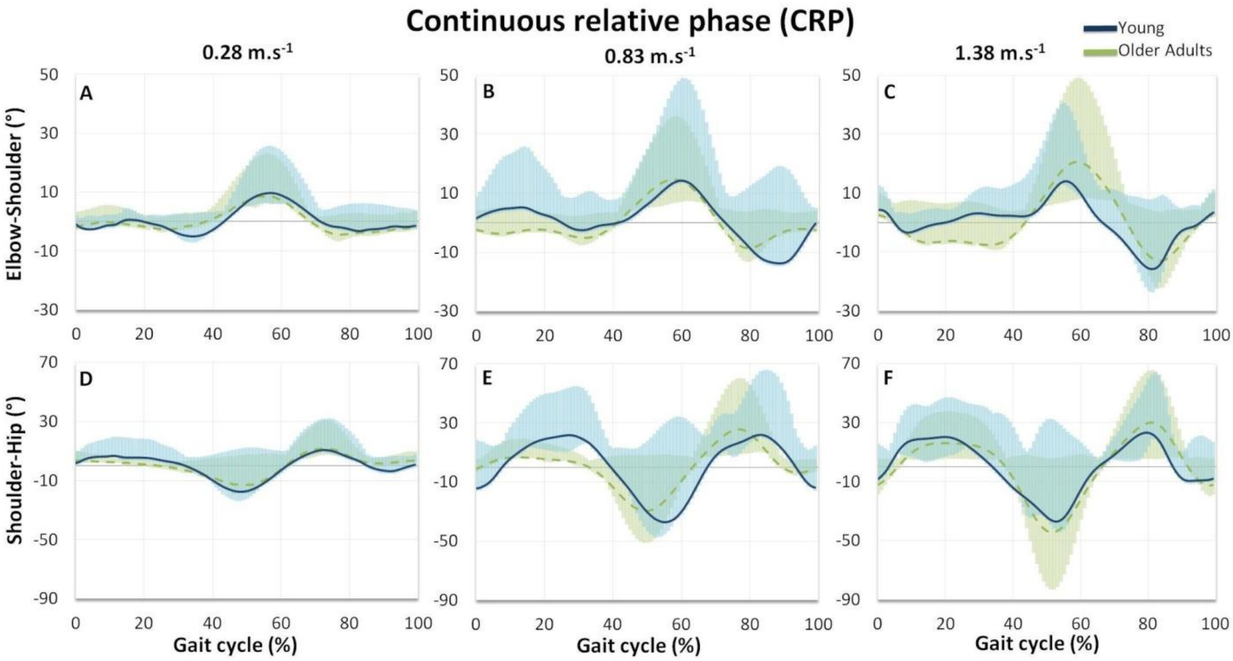
Gait Cycle Phase	Segment Couple	0.28 m.s^{-1}		0.83 m.s^{-1}		1.38 m.s^{-1}		Model Effect		
		Older adults	Young	Older adults	Young	Older adults	Young	Speed	Group	S*G
		Mean±SD	Mean±SD	Mean±SD	Mean±SD	Mean±SD	Mean±SD			
Total stance	Elbow-Shoulder	1.4±0.78	0.78±1.3^{b}	1.2±1.1	2.1±4.5	0.80±2.0†	4.4±3.6	0.071	0.040*	0.008*
	Shoulder-Hip	-3.4±1.5^{ab}	-2.5±2.4	-5.9±2.1†	-2.8±5.0	-5.7±2.1	-4.2±2.9	0.030*	0.005*	0.444
Loading response	Elbow-Shoulder	-0.75±4.7	-1.8±1.5	-3.2±4.5	1.7±7.7	-1.6±6.1	-0.53±2.3	0.935	0.114	0.109
	Shoulder-Hip	3.0±2.1^{b}	5.1±4.3	3.7±4.5$^{\dagger c}$	-4.9±14.3	-0.75±4.7†	6.6±8.5	0.091	0.867	0.006*
Mid-stance	Elbow-Shoulder	-1.9±1.6^{b}	-1.1±0.96	-3.2±2.9^{c}	-0.50±9.5	-6.6±6.4†	1.1±3.3	0.359	0.001*	<0.001*
	Shoulder-Hip	0.55±1.8^{ab}	4.6±6.3^{ab}	5.24±4.6$^{\dagger c}$	17.9±10.8	15.1±4.7	18.4±8.3	<0.001*	<0.001*	0.021*
Terminal stance	Elbow-Shoulder	2.0±3.3$^{\dagger b}$	-0.76±2.3	0.91±3.2	-0.83±7.3	-1.2±2.9†	3.7±6.5	0.658	0.923	0.007*
	Shoulder-Hip	-8.2±4.7^{a}	-9.5±6.4	-13.9±6.4†	-2.5±11.5	-11.0±5.4	-13.3±13.3	0.334	0.141	0.012*
Push-off	Elbow-Shoulder	7.8±5.5^{ab}	8.9±4.9	13.5±6.0^{c}	11.7±16.6	18.3±6.8†	11.6±9.0	0.001*	0.259	0.075
	Shoulder-Hip	-8.4±7.2^{ab}	-10.2±9.7^{ab}	-20.4±7.9^{c}	-34.0±27.2	-35.4±6.2	-36.3±29.1	<0.001*	0.152	0.381

Source: Author's own elaboration.

Note - Values are presented by means and standard deviation. Superscript symbols indicate statistically significant differences ($p<0.05$) in model effect (*), in groups within speeds (†), letters for the speeds within each group (a: 0.28-0.38 m.s^{-1}, b: 0.28-1.38 m.s^{-1}, c: 0.83-1.38 m.s^{-1}).

As expected, older adults had a reduced CRP and greater coordination stability at preferred and faster walking speeds (Figure 3.3.1). Velocity effects were observed in the final sub-phases of the joint pairs mainly for older adults (Table 3.3.2). We observed that young showed a linear increase in mean values, while the response of older adults was variable at the highest speed. For example, the elbow-shoulder CRP in total stance increased at three speeds for the young (0.78°, 2.2° and 4.4°) and decreased for the older adults (1.4°, 1.2° and 0.80°).

Figure 3.3.1 - Ensemble means elbow-shoulder and shoulder-hip CRP curves of older and young groups through a gait cycle.



Note - The solid blue line represents the average values of young individuals, the dash green line represents the older adults. The shaded lines represent the standard deviation of the means.

3.4 DISCUSSION

This study aimed to investigate age-related changes in intersegmental coordination and the coordination stability of upper extremity joints at different walking speeds. We confirmed our first hypothesis, showing that older adults have reduced CRP compared to young and there was a reduction with increasing speed (DEWOLF et al., 2019). Coordination stability, however, increased with aging refuting our second hypothesis. Our findings are in line with observations of Chiu; Chou (2012) and Hafer; Boyer (2018) that older adults walk with reduced CRP in lower limbs.

The intersegmental coordination of upper limbs showed a different behavior when comparing the older adults to the young. The older adults had higher values of CRP for intermediate speed and lower values for fast speed. However, when analyzing the movement synchronization, the older adults were more in phase than the young, regardless of speed, indicating a more cautious, more stable, and en-bloc gait of the older adults, in line with previous studies (IPPERSIEL; ROBBINS; DIXON, 2021). The older adults approached the values of 0º with increasing speed, contrary to the literature, which shows anti-phase coordination of the lower limbs (HAFER; BOYER, 2018, VAN EMMERIK, et al., 2005, HANG; DINGWELL, 2008).

Differently from young adults, there were differences in CRP between speeds in older adults, especially at slower speeds. Furthermore, most of the differences of the CRP between groups were at speeds of 0.83 and 1.38 $m.s^{-1}$. Chiu; Chou (2012) found these differences for the young but were not in older adults. Our results indicate that older adults adopt a strategy that reduces the control of balance at fast speeds, which may reflect impairments in the neuromuscular system.

The specific coordination pattern at mid-stance for elbow-shoulder and shoulder-hip were lower in older compared to young adults at intermediate and fast speeds. Hafer; Boyer (2018) observed that there is a specific attempt to maintain stability during single-leg loading with increasing age. Differences related to the speeds of the older adults for elbow-shoulder CRP were found at 0.38 - 1.38 $m.s^{-1}$ and 0.83 - 1.38 $m.s^{-1}$. Shoulder-hip CRP was different for the older adults at the three speeds. Furthermore, the joint combinations were anti-phase with increasing speed and age.

Studies show specific changes in coordination stability with aging (HAFER; BOYER, 2018, IPPERSIEL; ROBBINS; DIXON, 2021). Thus, confirming the results of this study. In addition, the coordination stability was different between groups at faster speeds. Corroborating, older adults seemed to maintain similar patterns in lower limb coordination when walking at different speeds, suggesting that they can modify the coordination pattern to sustain their adaptability when asked to walk faster or slower (HAFER; BOYER, 2018, IPPERSIEL; ROBBINS; DIXON, 2021). Although, on the other hand, Kang; Dingwell (2008) found lower coordination stability of the older adults' at slower speeds, likely most related to loss of strength and flexibility than to the speed effect.

Older adults were more coordinated during early stance and more variable during mid-swing on uneven surfaces (IPPERSIEL; ROBBINS; DIXON, 2021). We extend these findings specifically to upper limbs. When older adults are compared with younger ones, they show differences at faster speeds with a more in-phase gait. Aging causes shorter arm and waist swings. Caution strategies when walking may reflect an altered gait with healthy aging. Older adults use functional strategies differently for gait control compared to young people (HAFER; BOYER, 2018, IPPERSIEL; ROBBINS; DIXON, 2021). Gait training protocols as Nordic walking have shown improvements in functional mobility (GOMEÑUKA et al., 2019) and the oscillation of upper limbs (GOMUÑUKA et al., 2020). What still remains to be established is whether these gains in range of motion can be translated to a better upper limb intersegmental coordination. We did not include a fragilized older group, so our findings are limited to non-fragilized older adults.

In conclusion, older adults show altered upper limbs' CRP and coordination stability as compared to young and, these differences appear most often across at preferred and faster walking speeds. In addition, the speed affected the CRP of older adults, which increases the coordination stability of upper limbs from preferred to fast walking speeds. Therefore, our investigation provided insights related to changes in gait control with aging. We recommend that gait training protocols increase arm oscillation, aiming to enhance the intersegmental coordination and stability in older adults.

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3.6 SUPPLEMENTARY MATERIAL 1

Table 3.6.1 - Mean, standard deviation and statistical significance of for older adults and young during each functional period of the stance phase at different speeds.

Gait Cycle Phase	Segment Couple	0.28 m.s^{-1}		0.83 m.s^{-1}		1.38 m.s^{-1}		Model Effect		
		Older adults	Young	Older adults	Young	Older adults	Young	Speed	Group	S*G
		Mean±SD	Mean±SD	Mean±SD	Mean±SD	Mean±SD	Mean±SD			
Continuous relative phase angles										
Total stance	Hip-Knee	-0.81±1.3	-0.60±0.81	-1.0±1.5	2.0±11.9	-0.64±2.0	3.1±8.6	0.300	0.102	0.317
	Knee-Ankle	2.7±2.6^{ab}	2.5±2.4	6.6±2.6^{c}	7.6±6.8	8.0±2.3†	4.6±2.9	<0.001*	0.297	0.004*
	Pelvic-Thoracic	-0.50±0.43$^{\dagger ab}$	1.1±0.83	-1.1±0.72†	1.1±1.9	-0.97±0.75†	2.9±2.5	0.541	<0.001*	0.235
Loading response	Hip-Knee	1.3±4.3	2.1±6.9	7.8±3.1	1.1±18.1	9.6±3.5	8.4±10.2	0.008*	0.207	0.391
	Knee-Ankle	-0.30±4.5^{b}	-2.6±2.7^{a}	-1.0±3.0$^{\dagger c}$	16.9±18.8^{c}	4.9±4.5†	-0.49±6.7	<0.001*	0.097	<0.001*
	Pelvic-Thoracic	0.81±0.61^{ab}	-0.60±2.4	-0.12±0.91^{c}	-1.5±3.3	-1.6±1.2	-0.19±4.6	0.102	0.570	0.111
Mid-stance	Hip-Knee	1.8±2.1^{ab}	2.8±3.8	5.5±2.5^{c}	9.6±18.1	10.8±1.9	11.5±4.3	<0.001*	0.275	0.807
	Knee-Ankle	-1.5±4.2$^{\dagger b}$	-4.9±4.3^{b}	-3.0±2.4$^{\dagger c}$	-9.8±10.9	-10.9±4.9†	-14.4±4.1	<0.001*	0.002*	0.515
	Pelvic-Thoracic	-0.14±0.65$^{\dagger ab}$	1.2±1.7	-1.7±1.5†	1.3±3.5	-2.2±1.5†	3.3±2.0	0.350	<0.001*	0.001*
Terminal stance	Hip-Knee	0.20±2.8	-2.5±3.8^{a}	-0.1±4.6†	7.2±8.2	1.8±2.9	12.2±21.5	<0.001*	0.071	0.000*
	Knee-Ankle	4.9±3.7^{ab}	8.4±6.1	9.7±6.5†	2.6±8.0	7.8±4.7	4.5±6.4	0.956	0.081	0.026*
	Pelvic-Thoracic	-1.2±0.87†	2.3±1.5	-1.7±1.2$^{\dagger c}$	2.5±3.4	-0.85±1.5†	3.8±6.5	0.631	<0.001*	0.801
Push-off	Hip-Knee	-8.7±7.7^{ab}	-5.6±9.6^{b}	-21.7±11.7^{c}	-17.3±34.1	-33.0±7.8	-30.3±24.7	<0.001*	0.467	0.991
	Knee-Ankle	8.8±8.2^{ab}	9.7±12.8^{ab}	24.5±10.4^{c}	33.7±23.9	41.4±5.2	40.1±14.1	<0.001*	0.316	0.454
	Pelvic-Thoracic	-1.2±0.74$^{\dagger ab}$	0.55±1.6	-0.00±1.2^{c}	1.4±2.6	1.6±1.4	-0.12±4.8	0.022*	0.475	0.06
Continuous relative phase angle variability										
Total stance	Hip-Knee	2.7±5.6	9.0±16.3	0.92±2.6	11.1±20.6	1.5±2.7	7.8±13.0	0.832	0.066	0.772
	Knee-Ankle	7.6±7.4	8.3±1.2	9.9±2.3^{c}	15.4±14.7	11.2±2.1†	8.4±2.9	0.051	0.524	0.005*
	Pelvic-Thoracic	0.91±2.4$^{\dagger b}$	3.0±2.2	2.7±13.6	3.3±3.0	-0.3±0.77†	2.7±2.6	0.430	0.078	0.689
Loading response	Hip-Knee	3.8±6.3^{ab}	9.0±9.6	9.7±2.1^{c}	12.4±26.1	11.8±3.5	12.5±5.3	<0.001*	0.385	0.232
	Knee-Ankle	3.6±10.0	3.2±3.7^{a}	2.1±6.0$^{\dagger c}$	27.4±29.6^{c}	9.7±8.5†	3.6±6.3	0.012*	0.073	<0.001*
	Pelvic-Thoracic	2.1±2.2^{b}	1.6±3.6	2.3±8.0	0.60±3.5	-1.0±1.2	0.52±4.4	0.029*	0.855	0.253
Mid-stance	Hip-Knee	4.5±5.6^{b}	10.5±13.4	6.9±1.3^{c}	17.8±26.6	11.7±1.9	14.1±7.5	0.030*	0.081	0.408
	Knee-Ankle	2.0±8.5^{b}	0.90±4.7^{b}	-0.62±5.0^{c}	-2.8±20.2	-9.2±4.4	-11.7±5.2	<0.001*	0.457	0.852
	Pelvic-Thoracic	1.2±2.5^{b}	3.1±2.6	1.2±10.3	3.2±3.2	-1.7±1.5†	3.9±2.1	0.412	<0.001*	0.059
Terminal stance	Hip-Knee	3.6±4.9	8.0±20.5	2.0±4.6	13.5±15.2	4.0±2.9	18.3±31.9	0.516	0.070	0.310
	Knee-Ankle	10.2±8.3	13.7±5.1	12.7±5.5	8.8±13.8	10.6±4.8	7.9±8.2	0.386	0.563	0.259
	Pelvic-Thoracic	0.2±2.7†	4.0±1.1	2.2±14.2	4.6±3.9	-0.2±1.5†	4.4±6.5	0.709	0.005*	0.863
Push-off	Hip-Knee	-2.9±9.6^{ab}	8.1±22.5^{b}	-19.2±10.4^{c}	-4.6±47.3	-28.8±9.1	-23.2±19.4	<0.001*	0.129	0.450
	Knee-Ankle	16.1±8.4^{ab}	16.8±7.4^{ab}	30.0±8.7^{c}	42.4±24.7	46.1±6.3	45.9±10.1	<0.001*	0.142	0.256
	Pelvic-Thoracic	0.28±2.5$^{\dagger b}$	2.8±1.7	6.3±23.7	3.9±5.1	2.2±1.6	0.52±4.9	0.407	0.793	0.030*

Note - Values are presented by means and standard deviation. Superscript symbols indicate statistically significant differences ($p<0.05$) in model effect (*), in groups within speeds (†), letters for the speeds within each group (a: 0.28-0.38 m.s^{-1}, b: 0.28-1.38 m.s^{-1}, c: 0.83-1.38 m.s^{-1}).

3.6 SUPPLEMENTARY MATERIAL 2

Figure 3.6.1 - Figure representing the moment of kinematic collection.

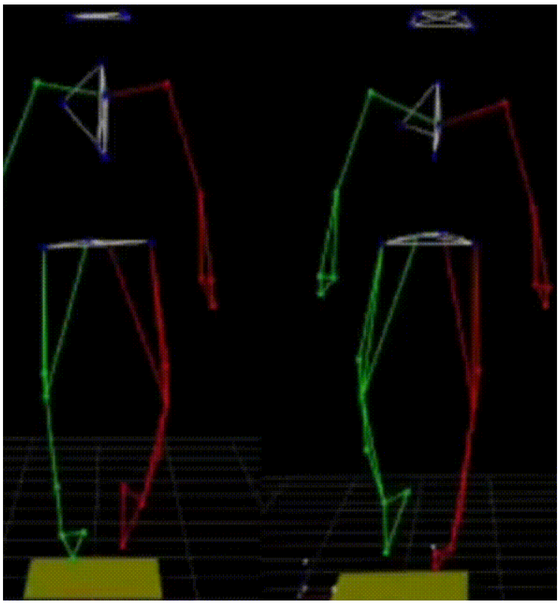


Figure 3.6.2 - Representative video of an older adult participant (left video) and a youth walking at a speed of 0.83 m.s^{-1}.

CHAPTER 4

STUDY 3

Bilateral coordination pattern is associated with interjoint coordination but not with variability during walking in older adults

4.1 INTRODUCTION

Daily living activities involve the ability to coordinate fundamental and complex movements. While the best effectiveness of these movements occurs in adulthood, deficits and lack of limb coordination occur due to aging (SWANSON; FLING, 2018). The aging process causes a decrease in functional capacity and changes in motor coordination can be explained by sarcopenia and dynapenia, loss of agility and balance, decrease range of motion and joint mobility, and greater rigidity in cartilage, tendons and ligaments (MITCHELL et al., 2012; STATHOKOSTAS et al., 2013; PAGAC, 2018). Walking is a specific task that incorporates many movements involving spatial and temporal coordination of the lower limbs, requiring control of the legs in time and space for effective and safe locomotion (HAUSDORFF et al., 2001; DEWOLF et al., 2019). Age-related changes also occur in the coordination of limb movements during walking, with some studies describing such changes comparing young and older adults (BOYER et al., 2017, CHIU; CHOU, 2012). Those studies have confirmed changes in coordination with age, with older adults having a motor pattern less responsive to disturbances and a greater stride-to-stride variability for stride width and length (BOYER et al., 2017, CHIU; CHOU, 2012; GRABINER et al., 2001).

Performing a stride during walking requires an ability to organize left and right step patterns, commonly known as gait coordination. Bilateral coordination represents a global phase space relation between body sides and can be quantified by the phase coordination index (PCI) (JAMES et al., 2016, GIMMON et al., 2018, PLOTNIK et al., 2007). PCI is a variable that integrates the bilateral accuracy (temporal similarity between contralateral steps) and consistency (stride-to-stride variability of bilateral accuracy, PLOTNIK et al., 2007). Plotnik et al. (2013) reported that aged adults have impaired bilateral coordination compared to young adults.

Such impairments in bilateral coordination could be more pronounced due to the age-related reductions in self-selected speed, which is probably mediated by reductions in step length (HAN et al., 2019).

Although the global variability (stride-to-stride fluctuations of the center of body mass) has been related to local variability (joint motion fluctuations), the relation between interjoint coordination and whole-body coordination has not been systematically evaluated. The interjoint coordination is commonly evaluated by the continuous relative phase analysis (CRP), which describes the space phase relation between adjacent joints. The variability of CRP represents the interjoint coordination stability or the resilience of the local motor system to disturbances (CHIU; CHOU, 2012; OGAYA et al., 2016; DEWOLF et al., 2019). As aging is associated with loss of mobility and impairments in gait mechanics, it would be of interest to use CRP to verify interjoint coordination and its variability in an older population. Older adults appear to alter the interjoint coordination in trunk (VAN EMMERIK et al., 2005), lower limbs (HAFER; BOYER, 2018), and upper limbs (MARTINS, submitted). Particularly, the foot-leg and leg-thigh coordination patterns in the final phase from stance are impaired and probably associated with age-related decline in step frequency (OGAYA et al., 2016). Older adults also reduce their range of lower limb motion resulting in altered interjoint coordination stability (DEWOLF et al., 2019). This strategy may be related to the control of hip and ankle movements during unipodal support since hip and ankle coordination was more affected than knee coordination (HAFER; BOYER, 2018).

Although interjoint and global coordination parameters (e.g., PCI) have been previously determined in aged people, it is still unknown whether interjoint coordination is related to bilateral coordination of gait in older adults. The main aim of this study was, therefore, to verify the relationship between bilateral coordination and interjoint coordination in able-bodied older adults during normal walking at different speeds. The secondary aim was to investigate relationships between bilateral consistency *versus* coordination stability, and bilateral coordination *versus* bilateral accuracy and bilateral consistency. We hypothesized that bilateral coordination could be explained by interjoint coordination. We also hypothesized that bilateral consistency would significantly and positively relate to coordination stability, while bilateral accuracy would be positively related to interjoint coordination.

4.2 MATERIALS AND METHODS

4.2.1 Study design

This study was observational, descriptive-exploratory, cross-sectional. All procedures were approved by the Ethics Committee of Research involving human beings from Universidade Federal do Rio Grande do Sul (project number: 33784014.7.0000.5347). All participants who agreed to participate signed an informed consent form.

4.2.2 Participants

Twenty-four healthy older adults (16 women) with a mean age of 66.3 ± 4.4 years, height of 1.65 ± 8.96 m and body mass of 78.2 ± 13.5 kg were recruited from the community. Participants were able to walk independently and had no history of falls in the last three months. The sample size calculation was performed based on the study Swanson; Fling (2018), which evaluated the effects of healthy aging on gait coordination. The calculation was performed in G*POWER version 3.1.6 software (Düsseldorf University, Düsseldorf, Germany) and considered the PCI effect size, adopting a significance level of 0.05 and a power of 95%. Inclusion criteria were related to good health, absence of neurological or cardiopulmonary problems and of history of neuromuscular, orthopedic, rheumatologic disorders or musculoskeletal deficiencies that prevented walking or performing the assessment (details in GOMEÑUKA et al., 2020).

4.2.3 Experimental protocol

All participants reported to the laboratory on two separate occasions over a 2-wk period. During the first visit to the laboratory, participants were familiarized with the testing procedures employed in the study and signed the informed consent form. After the preliminary visit, they returned to the laboratory to perform the walking test at different speeds on a treadmill.

The familiarization with the treadmill followed a standard protocol. The participants were introduced to the equipment and procedures and were also informed on the safety and support mechanisms of the research team. Afterward, the treadmill was started: each participant began walking at 0.13 $m.s^{-1}$, with a gradual adjustment of 0.13 $m.s^{-1}$ every minute until the individual could no longer walk, then starting the gradual deceleration. The preliminary visit lasted approximately 15 minutes.

During the treadmill walking test three-dimensional kinematic analyses were conducted using a motion capture system (Vicon Motion Capture System, Oxford, UK) with six infrared cameras (200 Hz, Bonita 1 MP, and T10 1.3 MP models). The system was calibrated according to the manufacturer's instructions for a space of 4 m (width), 6 m (length) and 2.5 m (height). All participants were instructed to wear dark sportswear and shoes suitable for physical activity, and thirty-five reflective spherical markers were placed on anatomical landmarks according to the Plug-in-Gait Full-Body model from NEXUS version 1.8.5 (Vicon Oxford, Oxford, United Kingdom).

After placing the reflective markers, each participant was positioned on the treadmill turned off (model ATL Inbrasport, Medgraphics, Ann Arbor, USA) to perform the static positioning collection for subsequent reconstruction. The participants performed one more familiarization with the treadmill (5-10 minutes) and were free to start data collection. They performed the treadmill walking tests for 5 minutes at each sub maximal walking speed (0.38, 0.83, and 1.38 $m.s^{-1}$) alternating with 5 minutes of rest, and the kinematic data collection was performed at the last minute of each speed.

4.2.4 Gait analysis

In order to focus on the quality of the walking assessment, the collected data were pre-processed and analyzed qualitatively from video recordings and temporal coordinate graphics. After pre-processing, the following temporal gait parameters were determined.

4.2.4.1 Data processing

The kinematic data were analyzed in a custom routine in Lab View version 2020 (National Instruments, Austin, USA). The raw curves of kinematic data were filtered with a low pass 7 Hz Butterworth 3º order filter. The stride events of touchdown and takeoff were determined by visual inspection in NEXUS software. One stride cycle was determined as the interval between two consecutive touchdowns from the same foot. The stride cycles and temporal variables were determined using the temporal events of touchdown and takeoff. The stride length was calculated by multiplying the treadmill speed by the stride cycle time.

The body segments vectors were determined using the markers' spatial position. The segment angular position was calculated using the segment inclination in relation to the global reference. The joint angular position was determined as the difference from the angular position of adjacent segments (WINTER, 2005). The angular velocity was calculated by the derivative of angular position in time.

4.2.4.2 Bilateral coordination - phase coordination index (PCI)

Bilateral coordination was calculated through PCI, which examines the accuracy of anti-phase coordination and gait pattern consistency. To our knowledge, PCI is the only metric to quantify whole-body coordination in terms of phase (SWANSON; FLING, 2018). The calculation of PCI followed the procedures proposed by Plotnik et al., (2007). Briefly, PCI is a metric that combines the accuracy and consistency of stepping phases generation with respect to the value of 180°. Lower PCI values reflect a more consistent and accurate phase generation and are related to different health conditions, while higher values indicate a more impaired bilateral coordination of gait. For the PCI calculation, the absolute deviation of the gait phase of 180° (bilateral accuracy) and the coefficient of variation of the gait phase (bilateral consistency) are necessary. A PCI value of 0, described as percent, indicates a "perfect" bilateral coordination, while values further away from 0 reflect increased impaired bilateral coordination (MEIJER et al., 2011, PLOTNIK et al., 2007; PLOTNIK et al., 2013).

Table 4.2.4.3.1 - Operational description of outcomes.

Outcome	Description
Bilateral coordination (%)	It is the phase coordination phase, calculated by the algebraic sum of bilateral consistency and accuracy.
Bilateral accuracy (%)	The similarity between contra lateral step times.
Bilateral consistency (%)	It means the variability of bilateral accuracy values across 10 strides.
Continuous relative phase (degree)	It means the interjoint coordination or the mean of instantaneous angular position of one joint relative with adjacent joint dynamically across different phases of stance.
Interjoint stability (%)	Represents the resilience to perturbation in interjoint terms and it is calculated as the variability of continuous relative phase across subsequent strides.

Source: Author's own elaboration.

4.2.5 Statistical analysis

Continuous variables were tested for normality using the Shapiro-Wilk test. To test our hypotheses, a Pearson correlation coefficient test (r) was initially performed. If the correlation was statistically significant, the r coefficient was classified as: very weak (≤ .19), weak (.20 to .39), moderate (.40 to .69), strong (.70 to .89), and very strong (≥ .90). In the present study, variables included for the correlation analysis were bilateral coordination, bilateral consistency, bilateral accuracy, interjoint coordination of hip-knee and knee-ankle across different phases of stance, and coordinative interjoint stability (see Table 4.2.4.3.1 for details). Then, a multiple regression model using the backward method was used to evaluate which independent variables, namely CRP and CRPsd, significantly influenced dependent variables (PCI, bilateral consistency, and bilateral accuracy). The principles of normality, multicollinearity, linearity, normality and independence of residuals, and homoscedasticity were respected. The level of significance for all analyses was $\alpha < .05$. The statistical tests were performed in SPSS® version 21.0 (IBM Corporation, USA).

4.3 RESULTS

Figure 4.3.1 generally describes the behavior of the CRP for a PCI that represents an impaired bilateral coordination (black point) and a near ideal coordination.

Figure 4.3.1 - Example angle-angle plots for the joint combinations of two older adults with different PCI. Black points represent an older adult with impaired bilateral coordination.

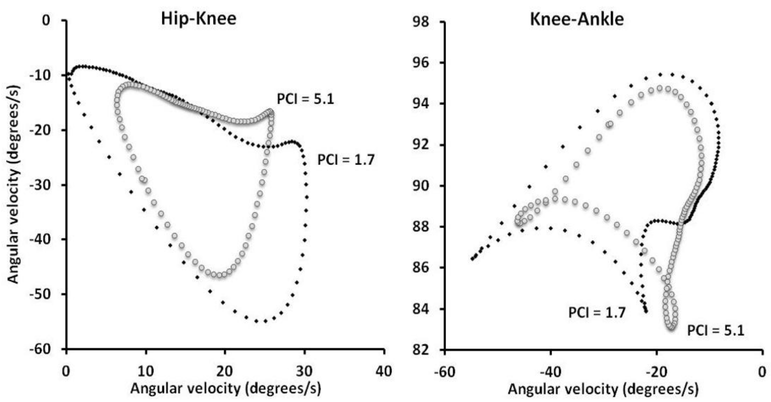


Table 4.3.1 shows the characteristics of the participants and the results of the bilateral coordination, interjoint coordination, and coordination stability.

Table 4.3.1 – Descriptive characteristics of the participants (n = 24).

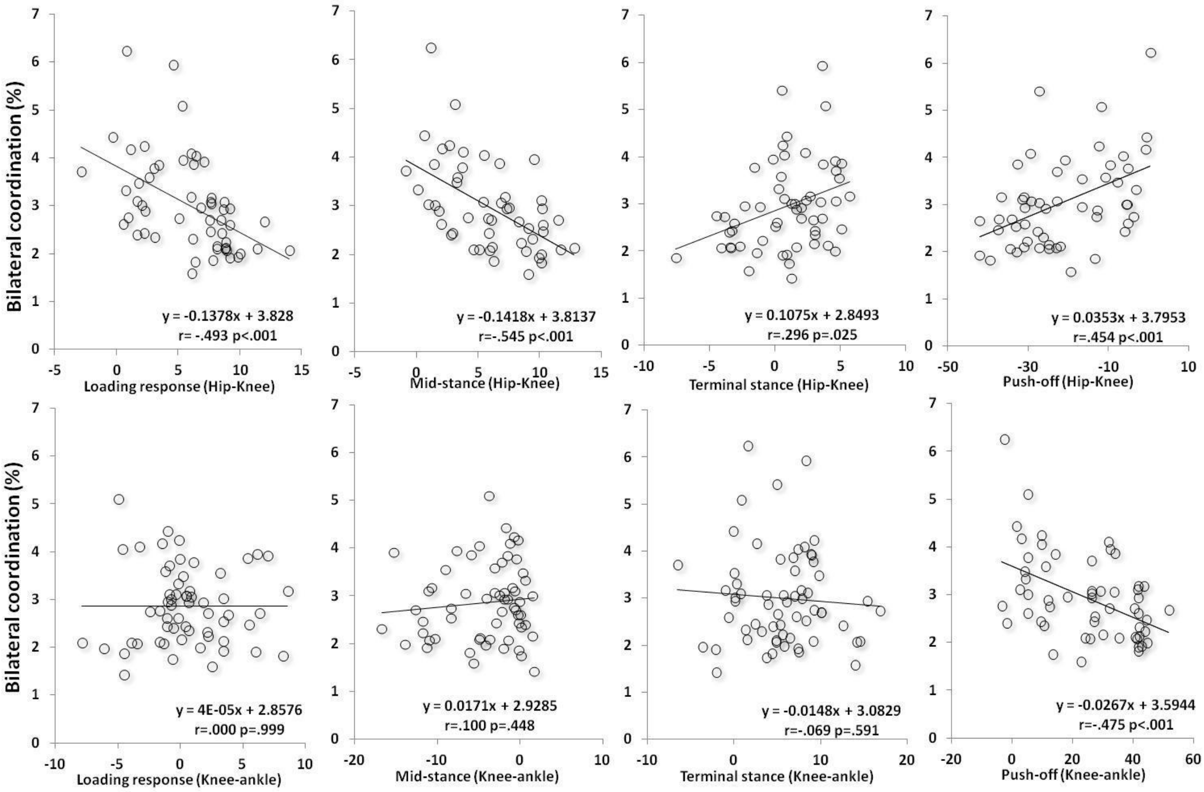
Variables		**Mean ± SD**
Characteristics	Age (years)	66.3 ± 4.3
	Height (cm)	165.2 ± 8.9
	Body mass (kg)	78.3 ± 13.5
Bilateral coordination	Phase coordination index (%)	3.0 ± 1.1
	Bilateral consistency (%)	1.51 ± 0.55
	Bilateral accuracy (°)	1.42 ± 0.56
	φ (°)	179.3 ± 1.9
Interjoint coordination Hip-knee continuous relative phase (º)	Loading response	5.8 ± 3.5
	Mid-stance	5.9 ± 3.5
	Terminal stance	.87 ± 2.9
	Push-off	-21.3 ± 12.2
	Coordination stability in loading response	-1.3 ± .8
	Coordination stability in mid-stance	-.9 ± .8
	Coordination stability in terminal stance	-2.0 ± 1.2
	Coordination stability in push-off	-3.7 ± 1.6
Interjoint coordination Knee-ankle continuous relative phase (º)	Loading response	.51 ± 3.4
	Mid-stance	-4.1 ± 4.7
	Terminal stance	5.5 ± 4.7
	Push-off	24.3 ± 15.9
	Coordination stability in loading response	-2.1 ± 1.4
	Coordination stability in mid-stance	-1.7 ± 1.2
	Coordination stability in terminal stance	-2.4 ± 1.5
	Coordination stability in push-off	-4.2 ± 2.7

Source: Author's own elaboration.

Note - SD: standard deviation; φ: stepping phase; phases of interjoint coordination - Loading response: angle at contact 0-20%, Mid-stance: angle at contact 20-50%, Terminal stance: angle at contact 50-80%, Push-off: angle at contact 80-100%.

The results of the correlation analysis are shown in figure 4.3.2, including the correlation coefficients of bilateral coordination with respect to interjoint coordination in each phase. Bilateral coordination was negatively correlated with hip-knee CRP in loading response ($r = -.493$, $p<.001$, moderate) and mid-stance ($r = -.545$, $p<.001$, moderate), but positively correlated with terminal stance ($r = .296$, $p=.025$, weak) and push-off ($r = .454$, $p<.001$, moderate). Bilateral coordination, otherwise, was negatively correlated with knee-ankle CRP only in the phase push-off ($r= -.475$, $p<.001$, moderate).

Figure 4.3.2 - Correlation analysis between bilateral coordination (PCI) and interjoint coordination (CRP) in each sub-phases of the support phase.



As noted in Figure 4.3.2, there is a relationship between local and global coordination. There is greater local variability for global adjustments for hip-knee CRP, thus indicating that one will show a more variable behavior than the other (minor variation). Bilateral coordination improved when interjoint coordination increased, showing an antiphase pattern, which characterizes a more robust gait. The knee-ankle CRP, otherwise, did not have a significant correlation with bilateral coordination in most phases, except in push-off, thus showing the antiphase pattern and improved coordination (minor bilateral coordination).

Table 4.3.2 shows the correlation coefficients between bilateral coordination and bilateral consistency with coordination stability. There were no significant correlations in all support sub-phases between bilateral coordination and coordination stability as well as between bilateral consistency and coordination stability.

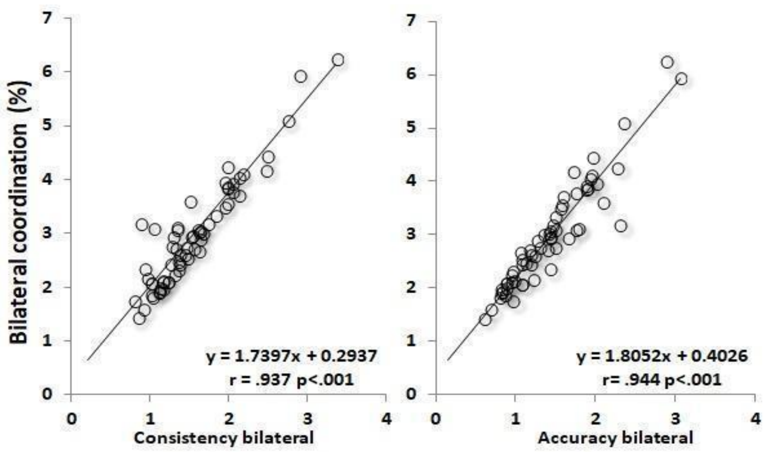
Table 4.3.2 - Correlation *r* coefficients between coordination variables and significance levels.

Variable		Bilateral coordination		Bilateral consistency	
		r	***p***	***r***	***p***
Hip-knee	Coordination stability in loading response	-.068	.623	.040	.777
	Coordination stability in mid-stance	-.168	.220	-.059	.669
	Coordination stability in terminal stance	-.069	.603	-.055	.681
	Coordination stability in push-off	-.133	.334	-.238	.083
Knee-ankle	Coordination stability in loading response	-.051	.713	-.011	.936
	Coordination stability in mid-stance	-.086	.529	-.048	.728
	Coordination stability in terminal stance	-.060	.659	-.025	.856
	Coordination stability in push-off	-.125	.350	-.241	.074

Source: Author's own elaboration.

The correlation coefficients of bilateral coordination with bilateral consistency and bilateral accuracy bilateral are shown in Figure 4.3.3. Bilateral coordination was positively correlated with bilateral consistency (r =.937, p<.001, very strong) and accuracy bilateral (r = .944, p<.001, very strong).

Figure 4.3.3 - Correlation analysis between bilateral coordination (PCI) and consistency bilateral and accuracy bilateral.



In order to follow the investigation for the relation between the dependent and independent variables, a multiple regression was conducted to verify whether interjoint coordination and coordination stability are capable of predicting bilateral coordination, bilateral consistency and bilateral accuracy (Table 4.3.3). The analysis resulted in a statistically significant model for the bilateral coordination [F(16.36)=2.430; p=.031; r^2=.660], bilateral consistency [F(16.35)=2.447; p=.030; r^2=.662] and bilateral accuracy [F(16.36)=2.182; p=.050; r^2=.636] (Table 4.3.3). The regression analysis estimated that 64-66% of the variance in bilateral coordination or bilateral consistency or bilateral accuracy could be explained by interjoint coordination.

The analyses of standardized regression coefficients (β; see Table 4.3.3) showed that the variance of bilateral coordination could be explained by hip-knee coordination stability in loading response (β=-.903; p=.018) and knee-ankle coordination stability in terminal stance (β=.749; p=.029). For bilateral consistency, the explicative variables were hip-knee coordination stability in push-off (β=-.548; p=.030) and knee-ankle coordination stability in terminal stance (β=.900; p=.010). Hip-knee coordination stability in loading response (β=-1.015; p=.011), otherwise, was the only explicative variable of bilateral accuracy.

Table 4.3.3 - Multiple regression models between bilateral coordination and interjoint coordination.

	r^2	p	df (F)	Adjusted β	t	p
Bilateral coordination	.660	.031	16.36(2.430)			
Hip-knee coord stability in loading response				-.903	-2.582	.018
Knee-ankle coord stability in terminal stance				.749	2.3591	.029
Bilateral consistency	.662	.030	16.36(2.447)			
Hip-knee coord stability in push-off				-.548	-2.332	.030
Knee-ankle coord stability in terminal stance				.900	2.842	.010
Bilateral accuracy	.636	.050	16.36(2.182)			
Hip-knee coord stability in loading response				-1.015	-2.803	.011

Note: coord - coordination; r^2 is the variance explained by the regression model. The degrees of freedom (df), F ratios, and p values are associated with the change in r^2. Only significant adjusted beta (β) coefficients are shown.

4.4 DISCUSSION

To the best of our knowledge, this study is the first to explore the principles underlying bilateral coordination and interjoint coordination and their potential interactions during walking in older adults. Two major findings emerged from this work. In line with our first hypothesis, interjoint coordination was significantly related to global coordination. The hypotheses relating the bilateral consistency and coordination stability, and relating the interjoint coordination and bilateral accuracy, however, were not confirmed. Hence, our results suggest that older adults present adjustments in interjoint coordination, particularly between hip and knee, when walking on a treadmill to maintain the whole-body motor control, with interjoint control playing a crucial role in bilateral coordination.

As previously noted, a negative correlation between bilateral coordination and interjoint coordination at the beginning of the support phase for proximal joints and the end of support phase for distal joints was found, indicating that CRP interfered with bilateral coordination. The current finding is of particular interest and may indicate that older adults that use a more anti-phase pattern of interjoint coordination walk with higher bilateral coordination. To the best of our knowledge, this is the first attempt to relate global and local coordination parameters of gait. We originally indicated that interjoint coordination accounts for bilateral coordination, with other plausible factors being observed as unrelated, such as gait asymmetry (PLOTNIK et al., 2007). The variation in bilateral coordination appeared to explain up to 66% by the lower limb coordinative stability. These results are in line with previous findings showing age-related impairments in bilateral coordination. In fact, our findings are similar to those reported by Plotnik et al. (2007), who found that the mean value of healthy young adults was 180.6°, while a mean value of 182.1° was found in older adults.

Previous findings have indicated a potential relationship of bilateral coordination with gait speed (PLOTNIK et al., 2007, GIMMON et al., 2018, HAN et al., 2019), which is in line with our present results. Also, bilateral coordination appears to be more impaired in older adults than in young adults, particularly at slow speeds (PLOTNIK et al., 2007). The changes in the sensory, neurological, or musculoskeletal systems caused by aging affect the motor tasks, causing impairments in posture and gait, reflecting a deterioration of balance and walking ability. This behavior was also observed in independent older adults (GIMMON et al., 2018), which can be explained by changes in gait characteristics and by the temporal gait parameters, in addition to being responsible for the long-term control of gait rhythmicity (HAUSDORFF et al., 2001; PLOTNIK et al., 2007; KIM et al., 2021). James et al. (2016) investigated the relationship between bilateral coordination and mobility and observed a significant association between these parameters (WINTER, 1995; CHIU; CHOI, 2012). From a functional point of view, the task of balance during walking seems to be controlled by the proximal joints and less by the distal joints, supporting our findings concerning interjoint coordination. Hence, it seems that the central nervous system determines strategies that modulate the motion in proximal joints to effectively impact the gait control (CHIU; CHOI, 2012).

The results of the present study extend the findings of Kim et al. (2021), where gait patterns in older adults, such as slower walking speed, longer double support time, and shorter single support time, contributed to increases in both gait variability and bilateral coordination. Furthermore, changes in bilateral coordination were also observed in previous studies using different tasks to walking. Bilateral coordination worsened when turning in large circles, further worsened during backward walking, and was worse when turning in small circles, where the highest shifts from anti-phase coordination were observed (PETERSON et al., 2012).

This study has some limitations. The current sample is relatively narrow in terms of medical condition and functional autonomy, with most participants being relatively healthy and independent. It is, therefore, difficult to generalize our findings to other populations, particularly for extremely weak or institutionalized older adults. The current sample is also narrow in terms of age. Prior studies investigating gait coordination parameters between different age groups in older cohorts have suggested particular alterations. Unfortunately, this hypothesis could not be tested in the context of the current study.

In summary, our results indicate a significant relationship between bilateral coordination and interjoint coordination in older adults. Specifically, older adults tend to present adjustments in interjoint coordination to improve bilateral coordination. The current findings suggest that interventions that enhance interjoint coordination (e.g. nordic walking or balance program) may increase bilateral gait coordination in aged people and improve their functional mobility.

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CHAPTER 5

INTEGRATIVE ANALYSIS OF RESULTS

5.1 GENERAL DISCUSSION

This thesis was organized into three articles that focused on the study of functional and coordination determinants of gait in older adults, and each of them was designed to answer specific questions. In the first study, we were able to show that aging causes declines in gait speeds and physical abilities. We observed that balance control may represent a hidden and relevant determinant of comfortable walking speed. In addition, we observed that LRI represents the conditioning capacities and WR represents the coordinative capacities ones, both variables resulting in precise clinical indicators for older adults. In the second study, we showed the differences between young and older adults for different coordinative parameters of the upper limbs during gait. Older adults have less intersegmental arm coordination than younger people. However, they have greater coordination stability. These differences show a reduction in balance control at slow speeds and, the changes in coordination stability emerge to support better adaptability. In the third study, we verified that there was a relationship between local and global coordination. Importantly, intersegmental coordination seems to affect the variance of bilateral coordination. Our seniors showed that a more anti-phasic pattern of intersegmental coordination provides a walk with better bilateral coordination.

The study of the gait of older adults is essential to understand the effects of the aging process, allowing individuals to age with preserved functional capacity and motor skills (PAGAC, 2018). Regarding functional capacity, it is intrinsically shaped at the biological level by the aging process itself. Specifically, the gradual loss of physiological reserves that will reflect on the development capacity and adaptability of individuals, indirectly influencing motor skills and gait. The literature shows that the spatiotemporal characteristics of gait are altered in normal aging and the assessment of these alterations can provide additional information on, for example, mobility dysfunction and the risk of falling in the older adults (ALMARWANI et al., 2016). Based on our results, it was possible to understand some gait characteristics that are influenced by aging and how they behave. For example, the study of motor skills that determine speed and step frequency, related functional mobility and maximum walking speed with walking performance parameters. Quantifying these relationships of possible impairment of gait is extremely important, especially to predict risk factors such as decreased functional capacity of the older adults (LAMOTH et al., 2011).

One of the findings of the second study in this thesis is related to gait speed. Age-associated declines in sensory and motor systems result in a more variable, cautious, and ‘en bloc’ gait pattern for the older adults. Symmetrical walks present a minimal energy expenditure, due to the exchange of mechanical energy of the center of mass with responses similar to an inverted pendulum (CAVAGNA et al., 1976; ALEXANDER, 2005). Studies show that gait speed, in addition to postural balance, were associated with a decline in basic and instrumental activities of daily living in older adults living in the community (VERMEULEN et al., 2011, NAKAMOTO et al., 2015, SLOOT et al., 2021). In addition, the aging process causes changes in the pattern of locomotion and neuromuscular control, resulting in an increase in metabolic cost (MIAN et al., 2006), which makes walking slower in the older adults, associated with biomechanical changes, weakness and muscle coactivation, and reduced range of motion. In addition to balance control, which represents a hidden determinant in the reduced speed pattern of the older adults.

From the analysis of the intersegmental coordination of the upper limbs, it was possible to unveil that the older adults adopt a strategy that reduces the control of balance and arm movement during walking at fast speeds, which may reflect impairments in the neuromuscular system, muscle, sensory and musculoskeletal function (HAFER; BOYER, 2018). A gait pattern that requires more attention requires more cognitive control and reduced subcortical automatism due to aging implies in a destabilizing effect (LAMOTH et al., 2011). Unlike young people, the older adults showed differences in intersegmental coordination and coordination stability between the tested speeds, especially at slower speeds. However, most of the differences between groups (older adults vs. young) were at faster speeds. Most importantly, the reduced gait speed in older adults is a compensatory mechanism used to maintain dynamical stability in healthy older adults. The second study reinforces the importance of investigating local variables to capture specific information about gait changes with aging, providing data on neuromuscular control during functional movement (IPPERSIEL; ROBBINS; DIXON, 2021). In addition to the importance of testing different walking speeds, as slower walking reveals age-related declines when compared to faster ones (ALMARWANI et al., 2016).

Global and local coordination patterns were studied in the third study. The main results show a deterministic relationship between the two, where the bilateral coordination patterns are associated with an intersegmental coordination, mainly in the hip-knee combination and only in the knee-ankle push-off. A more anti-phasic pattern provides walking with better bilateral coordination. However, age-related declines can affect the ability to plan and process interjoint coordination, causing greater difficulties in using different coordination patterns in response to changes in motor tasks (CHIU; CHOU, 2012).

We developed a conceptual model, based on our main results, to explain how conditioning and coordinating abilities affect the gait of the older adults (see figure 5.1). Motor coordination can influence coordination capacity, through body balance, while physical capacities influence mobility through conditioning capacities, with the aging process as an intrinsic determinant in all stages. The influence at each level is described below:

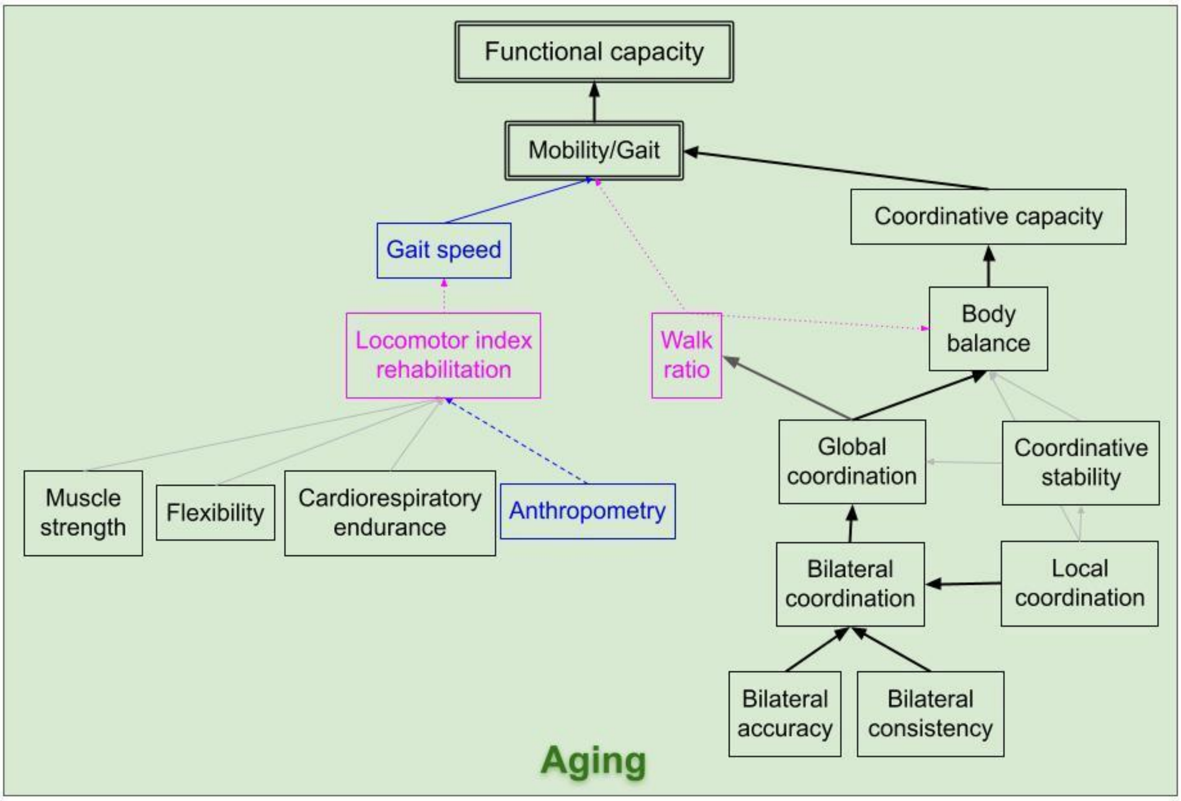

Figure 5.1 - Conceptual model of functional capacity and mobility in the older adults and its determinants from the aging process.

The model indicates clear evidence on separation of capabilities. While SSWS, LRI and WR are highly related to functional mobility, conditioning abilities seems to be more related to LRI, and coordination abilities are more related to WR. Gait is composed of numerous dynamic activities that require the integration of multiple sensory and motor pathways so that the central nervous system can coordinate the components involved (CASAL et al., 2021).

Walking is a fundamental skill for individuals along the entire life. Recognizing and understanding the specific gait characteristics during the stages of life can favor the understanding of the best care strategies for preventing or slowing down problems that will arise over the years. These findings show a possibility of intensifying the interpretations of physiomechanical models for the understanding of parameters and variables related to movement (specifically gait) and encouraging the application of these models in experimental studies with the older adults. Recently, the LRI has been proposed as an interesting marker to evaluate the functional mobility in adults (BARBOZA et al., 2021) because it has the benefit of controlling the size effect and estimating qualitatively how economically the individual is walking (PEYRÉ-TARTARUGA; MONTEIRO, 2016). The study 1 has shown that the LRI is a useful marker of functionality in older adults and the physical fitness seems to be related to LRI (and consequently to functional mobility). More studies applying randomized controlled trials are needed to confirm our findings. On the other hand, the WR is related to coordination abilities in older adults, confirming the potential of this simple variable as a screening analysis (ROTA et al., 2011). Importantly, the present thesis observed that upper limbs have specific adaptations in terms of interjoint coordination as observed previously in lower limbs (HAFER; BOYER, 2018; IPPERSIEL et al., 2021). An anti-phasic coordination pattern seems to be useful for older adults. Indeed, previous findings produced by our group have revealed that exercise interventions focusing on larger joint excursion implying in a more anti-phasic interjoint pattern in upper limbs as observed after Nordic training have improved the functional mobility (GOMEÑUKA et al., 2019) as well as have increased the range of shoulder and elbow joints resulting in a more coordinated motion pattern (GOMEÑUKA et al., 2020). Finally, the Thesis has shown that the local coordination pattern (interjoint coordination) has an impact on the global coordination showing specific adaptations in older adults. These findings, collectively, will contribute to health professionals better evaluating and prescribing interventions aiming to improve the functionality of older adults.

5.2 GENERAL CONCLUSION

Through the studies of this thesis, it was possible to quantitatively evaluate the functional and coordination parameters of gait, and, importantly, we found most functional and coordination determinants of functional mobility in older adults. Coordination determinants seem to be the interjoint coordination, not just from lower limbs but also from upper limbs, impacting the global coordination of gait and WR. Physical abilities can also determine the functional mobility in older adults, evaluated by LRI.

Most of our hypotheses were confirmed, as the aging process causes different changes in the older adults' gait patterns. On the other hand, some hypotheses could not be confirmed. We expected more relationships between the conditioning abilities and functional mobility. Furthermore, important relationships were revealed, mainly regarding global coordination. We observed that a smaller intersegmental coordination is related to an impaired global coordination in the older adults, which reveals a lower capacity to control movements.

E então, surge uma reflexão dentre as considerações finais

Como já relatado na apresentação desta Tese, este trabalho passou por uma grande reformulação devido à pandemia de COVID-19. Preciso salientar que o projeto "Marcha de idosos: Fisiomecânica e Funcionalidade" que tinha por objetivo compreender os efeitos de diferentes tipos de treinamento da aptidão física nos parâmetros cinemáticos e energéticos da marcha, além de indicadores de qualidade de vida em idosos com diferentes níveis de fragilidade está aprovado no comitê de ética desta universidade e na plataforma Brasil. O desejo é que um dia ele possa ser desenvolvido por algum estudante que tenha curiosidade sobre o assunto. Entretanto, o estudo que se formulou com a reestruturação desta tese oportuniza conhecimentos que podem ajudar na formulação e inovação das estratégias de proteção aos idosos, avaliações, metodologias e treinamentos físicos que podem auxiliar na prevenção e melhora de parâmetros clínicos e funcionais dos idosos. Defendo aqui as descobertas e conhecimentos importantes para a compreensão dos efeitos do envelhecimento sobre a marcha de indivíduos saudáveis, nossos resultados adicionam uma nova visão e complementam estudos anteriores. Além de lançar lacunas e possibilidades para novos estudos que envolvam a fisiomecânica da marcha, permitindo também que futuros pesquisadores selecionem metodologias apropriadas para estudos com os idosos.

O período pandêmico nos trouxe, e ainda traz diferentes ensinamentos... nossa realidade se transformou. Dentre os relacionados com a conclusão deste estudo, saliento a humildade, solidariedade e coragem. ***Humildade*** *para perceber que era preciso mudar e enfrentar as dificuldades impostas por um novo estudo em um pequeno período de tempo;* ***solidariedade*** *advinda do grupo que acolheu e se disponibilizou para ajudar em qualquer fase do estudo, o trabalho em equipe e a coletividade permanecerão e* ***coragem*** *para enfrentar os diferentes desafios de escrever um novo estudo em um período de tempo tão curto. Uma outra lição que me foi demonstrada tem a ver com o gerenciamento de tempo, quando usamos com inteligência as horas é possível ter sucesso nas tarefas. Claro que não posso deixar de salientar a importância do apoio institucional (fui bolsista CAPES) e o apoio familiar, principalmente do meu noivo, que contribuiu para que fosse possível focar nos meus objetivos.*

Com tudo isso reflito sobre as limitações que existiam, que foram impostas e que continuaram mesmo depois da conclusão deste estudo. Desenvolver uma Tese, além de toda a importância intrínseca que o ato carrega, é defender uma visão do pesquisador a partir das vivências oportunizadas pelo envolvimento com o mundo acadêmico. E foi a partir dessas vivências nos projetos de extensão e pesquisa da universidade que se chegou a estes problemas de pesquisa. Assim, assumo como uma potencial limitação o conhecimento prévio dos temas abordados. Com a mudança do estudo, novas limitações surgiram como o acesso ao banco de dados dos projetos e posterior generalização dos resultados.

Aproveito para fazer uma pequena reflexão sobre a experiência de vivenciar um doutorado. Vejo como uma experiência única e de real transformação como pesquisadora. Diferentes experiências me trouxeram até aqui e oportunizaram-me a escrita deste trabalho. Uma tarefa árdua para os momentos atuais, principalmente na situação que o país está enfrentando. Não quero aqui levantar nenhuma discussão, mas é preciso salientar a desvalorização da pesquisa e do desenvolvimento da ciência. Somente com resiliência e apoio do grande grupo de pesquisadores da universidade é possível desenvolver estudos importantes e relevantes para a comunidade e população brasileira.

Deixo aqui minha gratidão a todos os protagonistas desse grande ato de aprendizado e descobertas para se alcançar este título tão importante e desejado!

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CHAPTER 6

6.1 THE PUBLISHED STUDIES DURING DOCTOR'S DEGREE

All information can be verified at: Currículo Lattes Valeria Martins

6.1.1 Articles published in journals

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