Neuromechanics of walking in older adults

Influence of the physical activity level and trans-spinal cord direct current stimulation on the neuromechanical pattern of posture and locomotion of older and young adults

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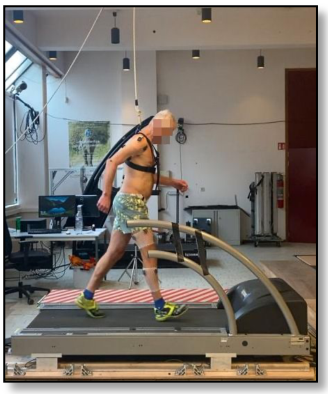

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Introduction

Preface

This chapter introduces the theoretical and empirical background underpinning the research presented in this thesis. A large part of the literature synthesis and conceptual framework outlined here has been published in the following peer-reviewed article Nùñez-Lisboa et al., 2023. Unraveling age-related impairment of the neuromuscular system: exploring biomechanical and neurophysiological perspectives. *Frontiers in Physiology*, 14. This paper is available as a Supplementary Chapter II (see page 163-175)

This publication was written during the initial phase of the PhD project to consolidate the existing knowledge in the field and identify key research gaps that the present thesis addresses. While some sections of the current introduction build upon that publication, the content has been revised and adapted to reflect the specific context, objectives, and scope of the thesis.

Defining Aging: A Multifaceted Process

Despite being a universal biological phenomenon, aging lacks a clear and universally accepted definition. Recent discussions have highlighted the ambiguity surrounding when aging begins, whether it should be classified as a disease, and how it should be measured or addressed (Mallapaty, 2024). From a biological perspective, aging has been described as the progressive accumulation of molecular and cellular damage that leads to a loss of physiological integrity, a concept formalized in the framework of the hallmarks of aging (López-Otín et al., 2013). Alternatively, the World Health Organization adopts a functional view, defining healthy aging as the process of developing and maintaining the functional ability that enables well-being in older age (World Health Organization, 2015). This perspective emphasizes not only intrinsic capacity but also the interaction with the environment. In this thesis, aging is operationally defined as a progressive decline in neuromuscular and sensorimotor function that compromises gait, balance, and autonomy. To examine how this decline may be mitigated through physical activity or neuromodulation, we included young (18–35 years) and older adults (≥65 years) of both sexes, all capable of walking independently.

Participants were classified as less or more active based on a median split of weekly physical activity, quantified in MET-minutes per week. This classification was derived from self-reported data using the Global Physical Activity Questionnaire (GPAQ), without applying predefined activity thresholds. In the tsDCS study, participants were randomly assigned to sham or stimulation groups, ensuring comparable physical activity levels between them to avoid baseline bias. Older adults were also screened using the Mini-Mental State Examination (MMSE ≥ 24) and were evaluated with the Short Physical Performance Battery (SPPB, score 3–9). Standard exclusion criteria were applied to reduce confounding, as detailed in Chapters III and IV.

From Mobility to Mechanisms: Why Gait Matters in Aging

Walking is a fundamental motor behavior that defines human mobility (Herman et al., 2005; Hollman et al., 2011). It is an essential activity of daily living, enabling individuals to interact with their environment, maintain independence, and participate in social and community life (Studenski, 2011). Unlike other repetitive tasks, walking requires the continuous coordination of sensory input, muscle control, and neural integration (Takakusaki, 2017). Its execution depends not only on muscular strength but also on the precise timing and regulation of movements across multiple joints and body segments.

In older adults, walking acquires even greater significance. As physical and neural capacities decline with age, the ability to walk safely and efficiently becomes a central component of functional autonomy (Abellan Van Kan et al., 2009; Montero-Odasso et al., 2012). Gait impairments are strongly associated with reduced quality of life, increased risk of falls, hospitalization, and long-term care admission (Studenski, 2011; Verghese et al., 2009). For this reason, gait is often considered a vital sign of aging, providing a sensitive indicator of overall health and neuromuscular function (Montero-Odasso et al., 2012). These observations highlight the need to examine the physiological mechanisms underlying gait control in later life, with particular attention to the neuromuscular decline that accompanies aging.

Life expectancy is steadily increasing, but maintaining quality of life in older age remains a major challenge (Abell et al., 2020). While people live longer, their ability to move independently often declines, compromising autonomy and well-being (Visser & Schaap, 2011). Among the biological systems affected by aging, the neuromuscular system plays a central role in gait decline. Its progressive deterioration, including motor unit loss, impaired neuromuscular junctions, and reduced sensory feedback, compromises both mobility and motor control (Cruz-Jentoft et al., 2019; Hepple & Rice, 2016; Rygiel et al., 2016a; Wilkinson et al., 2018). These changes disrupt movement coordination and reduce adaptability, leading to inefficient and unstable gait. They also provide the rationale for exploring interventions such as physical activity or tsDCS that aim to preserve or restore neuromechanical function.

Mechanical Cost and Transition Strategies in Aging Gait

A core feature of human walking is the capacity to regulate mechanical energy efficiently across steps. This efficiency depends on the inverted pendulum-like behavior of the center of mass (CoM), which rises and falls over the stance leg, allowing the exchange between kinetic (E_k) and potential energy (E_p) (Cavagna et al., 1977). In young adults, this passive mechanism minimizes energy cost. However, aging alters this dynamic. Older adults often display a higher external mechanical work (W_{ext}) to move the CoM, even at comparable walking speeds (Dewolf et al., 2022), indicating a reduced capacity to recover energy between steps.

These mechanical changes are accompanied by a redistribution of joint contributions to propulsion. With age, there is a well-documented shift from ankle push-off to hip-based propulsion, a phenomenon known as the distal-to-proximal redistribution of joint work (DeVita & Hortobagyi, 2000; Franz, 2016). This adaptation is frequently interpreted as a compensatory response to reduced distal power output, especially at the ankle (Delabastita et al., 2021; Franz, 2016), a pattern consistent with the cautious gait described by Boyer et al., (2023). This phenomenon, known as the distal-to-proximal redistribution of joint work, is one of the most robustly documented features of aging gait (DeVita & Hortobagyi, 2000; Franz, 2016; Waanders et al., 2019). Interestingly, this mechanical adaptation often coincides with changes in the timing of step-to-step transitions. Some older adults show a delayed redirection of the CoM following foot contact, a pattern defined as a non-anticipated transition, while others maintain a more timely redirection, referred to as an anticipated transition (Chastan et al., 2008; Dewolf et al., 2022; Meurisse et al., 2019b). These temporal strategies may reflect individual differences in how propulsion and stability are managed with aging.

Taken together, these findings suggest that age-related gait changes involve both increased mechanical cost and altered coordination during the step-to-step transition. As shown in Fig. 0.1, older adults exhibit higher W_{ext} and lower energy recovery compared to young adults, with substantial variability among individuals. This variability appears to align with distinct transition strategies, previously described as anticipated and non-anticipated. These differences may reflect individual adaptations in motor control that affect the efficiency of CoM redirection and overall gait performance.

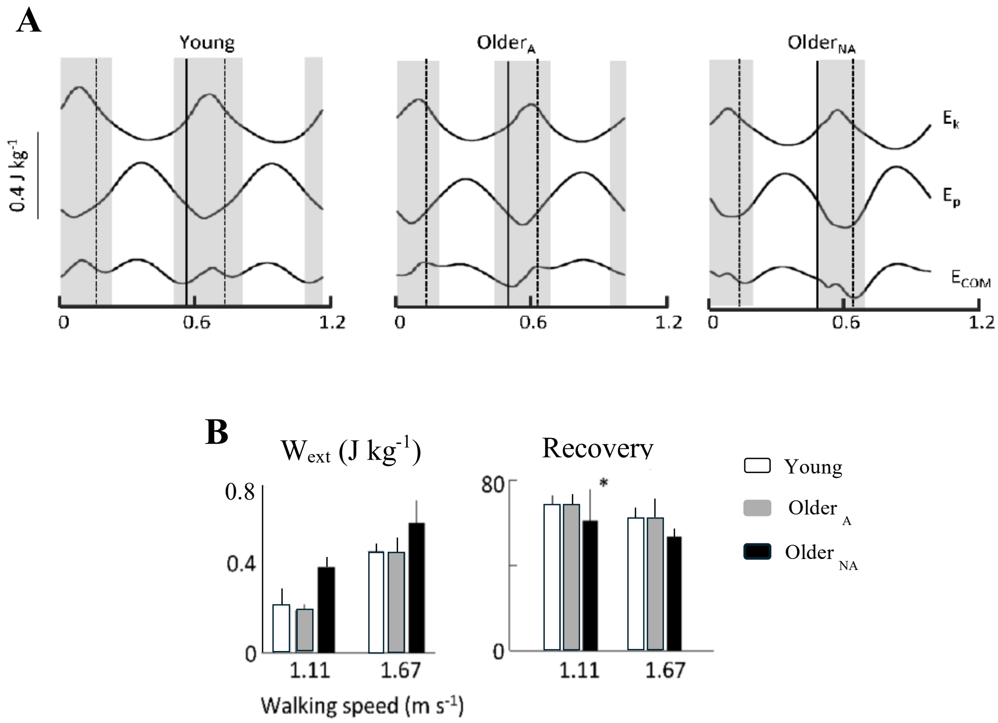

Fig. 0.1. *External work and energy recovery across groups. (**A**) CoM energy profiles (E_k, E_p, E_{COM}) in young adults, older adults with anticipated ($Older_A$), and non-anticipated ($Older_{NA}$) transitions. The gray shaded areas represent the double support phase, defined between the minimum and maximum vertical velocity of the CoM. The solid vertical black line indicates foot contact, and the dashed vertical black line indicates toe off. (**B**) Comparative bar graphs show higher W_{ext} and lower % recovery in $Older_{NA}$. Adapted from Dewolf et al. (2022).*

In summary, aging is associated with increased mechanical cost and altered coordination during walking. The presence of distinct transition patterns suggests individual differences in motor control strategy. Understanding these differences requires examining the neural mechanisms that support gait regulation.

Spinal and Neuromuscular Reorganization of Gait Control in Aging and Inactivity

While biomechanical adaptations are a hallmark of aging gait, they do not fully account for the underlying mechanisms of motor decline. Emerging evidence indicates that aging also affects the neural systems involved in gait regulation, including changes in motor coordination and spinal control. These neurophysiological changes contribute to altered gait patterns and compromised mobility in later life. Gait disturbances are among the earliest and most functionally limiting signs of neuromuscular aging, reflecting both muscular impairment and compensatory shifts in motor strategy.

Although age-related gait deterioration has been traditionally attributed to declines in muscle strength and peripheral function (Prince et al., 1997). Reduced mobility is highly prevalent among older adults and contributes to the progressive loss of muscle mass and strength, a condition known as sarcopenia (Landi et al., 2012). Notably, sedentary behavior has been associated with impaired reinnervation following motor unit loss, exacerbating the decline in neuromuscular function (Jones et al., 2022). In addition to sarcopenia, aging leads to structural changes at the neuromuscular junction (NMJ), reduced motor unit recruitment, and altered joint mechanics (Cruz-Jentoft et al., 2019; Power et al., 2010), all of which impair force generation and postural stability.

These physiological changes are likely to affect the neural control of walking, potentially in older adults with lower physical activity levels. In humans, walking is regulated in part by spinal motor neurons organized along the lumbar and sacral segments of the spinal cord (Ivanenko et al., 2006; F. P. Kendall et al., 2005). These segments innervate different groups of lower limb muscles. The lumbar segments (L2 to L5) predominantly control proximal muscles, including extensors such as the quadriceps, as well as muscles like the biceps femoris. In contrast, the sacral segments (S1 to S2) are more strongly associated with distal muscles, such as the gastrocnemius, soleus, and tibialis anterior, which contribute to push-off and foot clearance during walking. Understanding this anatomical distribution is essential for interpreting how aging affects motor control during locomotion (Cappellini et al., 2010; F. P. Kendall et al., 2005; Monaco et al., 2010). This segmental distribution reflects a rostro-caudal organization of spinal motor control, originating in the lumbar and extending into the sacral segments of the spinal cord. Motor pools are arranged along this axis, with each segment contributing to the coordinated activation of specific lower-limb muscles throughout the gait cycle (Avaltroni et al., 2024).

To illustrate this organization, Table 0.1 summarizes the spinal innervation of key lower-limb muscles based on reference segmental charts compiled from anatomical and clinical sources (Dewolf et al., 2021b; Kendall et al., 2005b).

Table 0.1 Innervation of the lower limb muscles

	Gmax	**TFL**	**BF**	**ST**	**VM**	**VL**	**RF**	**TA**	**PERL**	**GM**	**GL**	**SOL**
L2					X	X	X					
L3					X	X	X					
L4		X		x	X	X	X	X	x			
L5	X	X	x	X				X	X			x
S1	X	X	X	X				X	X	X	X	X
S2	X		X	X						X	X	X

Reference segmental charts for lower limb muscles from Kendall et al. (2005), obtained by combining the anatomical and clinical data from six different sources. A capital X denotes localization agreed on by five or more sources; a small x denotes agreement of three to four ssiources. Muscle abbreviations: Gmax = Gluteus maximus, TFL = Tensor fasciae latae, BF

= Biceps femoris, ST = Semitendinosus, VM = Vastus medialis, VL = Vastus lateralis, RF = Rectus femoris, TA = Tibialis anterior, PERL = Peroneus longus, GM = Gastrocnemius medialis, GL = Gastrocnemius lateralis, SOL = Soleus.

With aging, this rostro-caudal pattern becomes less distinct. Neural control of walking shows signs of reorganization, particularly in distal muscles innervated by sacral segments (Dewolf et al., 2021b). The activity of muscles such as the gastrocnemius, soleus, and tibialis anterior tends to become broader and more persistent throughout the gait cycle, especially around the push-off phase. In contrast, muscles with dominant lumbar innervation tend to show more stable activation profiles, suggesting that age-related adaptations are more prominent in circuits controlling distal muscles (Dewolf et al., 2021b; Monaco et al., 2010).

Compared to younger adults, older individuals often display a more diffuse and prolonged motor output from sacral spinal segments (see Fig. 0.2). This has been interpreted as a compensatory mechanism aimed at preserving forward propulsion and balance in the presence of declining distal neuromuscular performance. In this context, the increased reliance on sacral circuits reflects a shift in control strategy, helping to maintain gait despite age-related impairments (Avaltroni et al., 2024; Dewolf et al., 2021b).

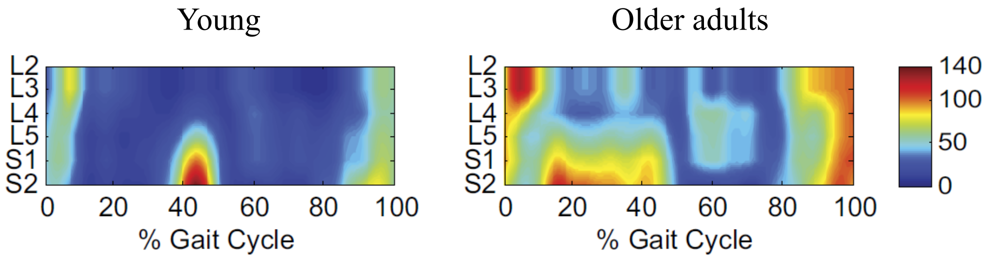

Fig. 0.2. *Segmental motor output during the gait cycle in young and older adults. Older adults show a more diffuse and prolonged activation across lumbar and sacral segments compared to young adults. Measurements were taken during treadmill walking at a fixed cadence of 120 steps per minute. Adapted from Monaco et al. (2010).*

These segmental adaptations are mirrored by temporal shifts in muscle activation patterns, particularly in muscles involved in push-off and postural support (Dewolf et al., 2021a, 2021b; Monaco et al., 2010). Rather than reflecting dysfunction alone, these patterns may indicate a reorganization of motor control aimed at maintaining dynamic stability. Interestingly, similar features have been described during early motor development, suggesting that such reorganization in aging could also be interpreted through the lens of developmental motor control.

Developmental and Aging Parallels in Gait

The segmental motor output shown above also reveals a striking similarity between the two ends of the lifespan. Both toddlers and older adults tend to exhibit broader and less differentiated spinal activation patterns, especially at the sacral level, reflecting a loss of rostro-caudal organization and reduced motor selectivity (see Fig. 0.3) (A. H. Dewolf et al., 2020; Ivanenko et al., 2013).

Functionally, this diffuse spinal output has been linked to a non-anticipated transition strategy during double support, characterized by delayed CoM redirection and increased mechanical cost in older adults (Dewolf et al., 2022). While in toddlers this pattern is part of normal gait development, in older adults it may reflect compensatory adaptation to neuromuscular decline (Dewolf et al., 2020, 2022). These lifespan similarities raise the hypothesis that gait control may rely on shared neuromechanical strategies under reduced corticospinal modulation. This hypothesis is explored in detail in the Supplementary Chapter II of this thesis (section: *Gait during development and aging: a brief overview of the two sides of life*), where segmental activation and transition strategies are compared across age groups.

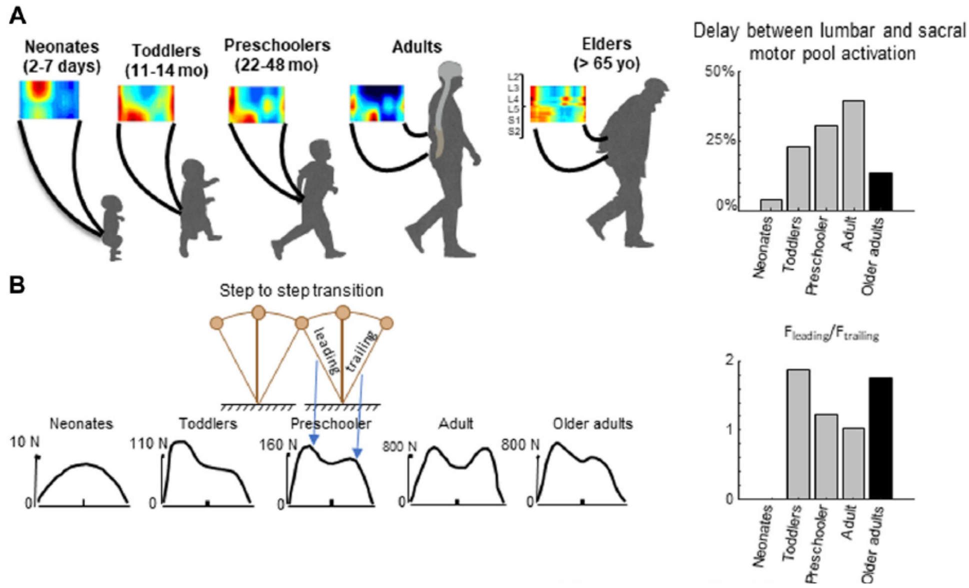

Fig. 0.3. *Lifespan comparison of spinal motor output and step-to-step transition mechanics.**(A)** illustrates segmental activation profiles from neonates to older adults, highlighting a diffuse and overlapping lumbar-sacral pattern in early and late life. The bar graph shows the progressive increase, and later decrease, of the delay between lumbar and sacral activation. **(B)** shows vertical ground reaction force patterns and step-to-step transition mechanics. Both toddlers and older adults exhibit increased mechanical contribution of the leading limb and diminished push-off from the trailing limb. Adapted from Núñez-Lisboa et al. (2023).*

The similarities observed between early development and aging suggest that non-anticipated transitions may emerge under conditions of reduced neuromechanical control, whether due to immaturity or decline. To provide a developmental contrast to the aging-related gait patterns described in this thesis, a Supplementary Chapter I entitled *"Effect of age and speed on the step-to-step transition strategies in children"* was included. This analysis investigates how transition strategies emerge and evolve during early motor development. By characterizing the maturation of CoM redirection and step-to-step coordination in children, it offers a developmental reference that supports the interpretation of transition impairments observed in older adults.

Postural Modulation of Neuromechanical Gait

Although age-related gait changes are often attributed to neuromuscular decline, posture may independently constrain the neuromechanical efficiency of walking. Anterior trunk inclination, commonly observed in older adults (Honda et al., 2023), alters the alignment of the CoM and may compromise both mechanical and muscular coordination during gait. Experimental studies suggest that artificially increasing trunk flexion in healthy young adults elevates W_{ext} and disrupts the timing and magnitude of muscle activation (Alghamdi & Preece, 2020; Grasso et al., 2000). These changes resemble neuromuscular patterns commonly observed in pathological gait, including altered timing and magnitude of muscle activation (Martino et al., 2014). This raises the hypothesis that trunk orientation, regardless of age, may act as a primary modulator of neuromechanical function during walking, a possibility further explored in Chapter II.

Approaches to Mitigate Age-Related Gait Impairments

In this thesis, we investigate two complementary strategies to counteract age-related gait decline. The first focuses on the role of physical activity level in modulating neuromechanical gait control, assessing how habitual exercise habits influence step-to-step transitions and energetic efficiency. The second strategy explores the potential of trans-spinal direct current stimulation (tsDCS) as a non-invasive intervention to enhance spinal motor output and improve locomotor performance. By combining observational and interventional approaches, this work aims to provide new insights into how behavioral and neuromodulatory factors can mitigate the functional consequences of neuromuscular aging.

The focus on physical activity is grounded in evidence that inactivity accelerates neuromuscular decline in aging, contributing to motor unit loss, impaired reinnervation, and reduced muscle function (Jones et al., 2022; Wilkinson et al., 2018). McPhee et al., (2016) further report that the absence of regular physical activity in older adults is associated with a major decline in multiple physiological systems, including reductions in muscle mass and strength, bone mineral density, cardiorespiratory, metabolic function, as well as neuromuscular impairments. Conversely, lifelong physical activity appears to preserve neuromuscular integrity by maintaining MU (motor unit) function and reducing neuromuscular junction (NMJ) dysfunction, potentially mitigating age-related declines in muscle activation (Power et al., 2010; Wilkinson et al., 2018). Moreover, exercise promotes MU plasticity by facilitating the reinnervation of denervated muscle fibers, a key mechanism for sustaining muscle function (Mosole et al., 2014). This neurophysiological adaptation supports autonomy and functional capacity, particularly in locomotion, and extends to gait control as aging is associated with altered movement patterns and neuromechanical adjustments. In this broader perspective, Lazarus & Harridge, (2018) concluded that lifelong engagement in exercise preserves neurophysiological integrity and represents a fundamental determinant of optimal aging.

Given these neuromuscular benefits, targeted exercise interventions have been widely explored as a strategy to counteract the functional decline associated with aging. Resistance training (RT) and other forms of structured exercise are widely recognized for their role in preserving muscular function in older adults (Chen et al., 2021; El Hadouchi et al., 2022; Kraemer, Ratamess, et al., 2002; Markov et al., 2023). Given its broad health benefits, current guidelines emphasize the importance of maintaining high levels of physical activity, as regular exercise is expected to enhance functional performance, particularly in gait (Boyer et al., 2012; D. Taylor, 2014). In alignment with these recommendations, research has consistently shown that RT and power training contribute to improvements in walking speed (Beijersbergen, Granacher, Gäbler, DeVita, et al., 2017; Beijersbergen, Granacher, Gäbler, Devita, et al., 2017a, 2017b; Hortobágyi et al., 2015; Stock et al., 2019). However, despite these functional gains, training does not necessarily translate into increased propulsive power generation during walking (Beijersbergen et al., 2013). While the biomechanical mechanisms underlying these adaptations remain unclear, the observed improvements in functional markers such as walking speed are well-documented. At the same time, the limited impact on propulsive power highlights a gap in understanding how physical activity levels influence gait neuromechanics.

Another non-invasive strategy proposed to enhance spinal motor control is tsDCS. This technique applies low-intensity electrical currents over the spinal cord and has been shown to modulate spinal excitability and reflex pathways (Awosika et al., 2019; W. Song & Martin, 2017). Evidence from healthy adults and clinical populations suggests that tsDCS can influence motor output, improve reflex modulation, with the most consistent improvements observed in walking speed and jump capacity (Awosika et al., 2019; Baczyk, M., 2019; Berry et al., 2017; Jankowska, 2017). Although tsDCS has not yet been tested in older adult populations, its documented effects on spinal circuitry support its potential as a neuromodulatory intervention to counteract gait impairments associated with neuromuscular aging. In general terms, anodal tsDCS tends to bias spinal circuits toward facilitation, likely through small shifts in segmental gain and motoneuron depolarization, whereas cathodal tends to reduce excitability; putative mechanisms include subtle reweighting of descending drive, ascending proprioceptive and cutaneous input, and local inter-neuronal circuits such as central pattern generating (CPG) and premotor networks. In this thesis, tsDCS is explored as a complementary strategy to physical activity for preserving locomotor function in aging.

Scientific and Social Relevance

The loss of mobility in older adults represents a pressing public health concern, with far-reaching implications at both individual and societal levels. As life expectancy increases, preserving autonomy and quality of life becomes a major priority (Freiberger et al., 2020). However, age-related mobility impairments dramatically elevate the risk of falls, which are a leading cause of injury, hospitalization, and long-term disability in the elderly (Blake et al., 1988; Robinovitch et al., 2013; Vaishya & Vaish, 2020). These events not only compromise physical independence but also trigger fear of falling, reduced social participation, and progressive functional decline (Pin & Spini, 2016). Falls and related injuries represent a substantial economic burden, accounting for billions in healthcare expenditures and resource allocation (Hartholt et al., 2011; Vaishya & Vaish, 2020). Moreover, mobility limitations are strongly associated with increased mortality, institutionalization, and psychological distress (Freiberger et al., 2020; Pin & Spini, 2016). Thus, understanding and preventing gait dysfunction and falls is essential not only to promote healthy aging but also to reduce the broader societal impact of age-related functional decline. Therefore, it is crucial to investigate how physical fitness levels and non-invasive neuromodulatory interventions, such as tsDCS, may contribute to improving gait function and mitigating mobility decline in older adults. This rationale underpins the central hypothesis and objectives of the present thesis.

In addition, the similarities observed between older adults and young children in neuromuscular control and gait organization suggest an opportunity for translational application. While physical activity interventions are already established in pediatric populations, non-invasive neuromodulation remains largely unexplored outside of clinical settings such as cerebral palsy. Demonstrating the efficacy of tsDCS in older adults could therefore provide a compelling foundation for future studies targeting children with reduced physical activity, such as those with obesity or developmental coordination disorders. This perspective reinforces the broader relevance of the present thesis across the lifespan.

Research objectives

This thesis aims to explore how aging, physical activity level, and spinal neuromodulation influence gait control in humans. Specifically, it examines how these factors affect neuromechanical parameters such as movement coordination, mechanical cost, and motor output organization during walking.

To address these aims, the thesis is organized into five main chapters. Chapter I outlines the general methodology, detailing the experimental setup, signal acquisition procedures, and data processing strategies employed across studies. Chapter II focuses on the biomechanical and neuromuscular consequences of anterior trunk inclination during gait. Chapter III investigates the role of habitual physical activity in modulating age-related changes in gait control. Chapter IV evaluates the acute effects of tsDCS on spinal motor output and postural alignment during walking. Finally, Chapter V presents a general discussion that integrates the experimental findings and contextualizes them within broader neuromechanical and functional frameworks of aging.

In addition to the core chapters, two supplementary chapters are included. The first explores the development of step-to-step transition strategies in children, offering a lifespan perspective on gait control mechanisms. The second compiles foundational concepts and empirical findings published earlier in the PhD project, which helped shape the conceptual framework of the present thesis. Together, these components provide a comprehensive investigation of the neuromechanical factors underlying gait decline with age and explore potential strategies to preserve or enhance locomotor function.

Hypotheses

Based on the conceptual framework presented in the introduction, the present thesis is structured around three central hypotheses.

First, that anterior trunk flexion alters step-to-step transition mechanics and increases the mechanical cost of CoM displacement, thereby providing a mechanistic basis to interpret age-related differences in locomotor control.

Second, that the level of habitual physical activity modulates age-related neuromechanical adaptations during gait, such that less active older adults exhibit greater mechanical work and cost of the CoM and altered rostro-caudal patterns of muscle activation compared with more active older adults and with younger adults.

Third, that an acute application of tsDCS older adults reduces the mechanical work and cost of the CoM and reorganizes the rostro-caudal activation pattern toward that observed in younger adults.

Specific Aims

To examine the impact of anterior trunk flexion on step-to-step transition strategies and CoM mechanics, providing mechanistic insight into the determinants of gait.

To analyze the influence of habitual physical activity level on age-related adaptations in gait, focusing on step-to-step transition mechanics, CoM mechanical work and cost, and rostro–caudal activation patterns in older and younger adults.

To determine the effects of an acute application of tsDCS in older adults on CoM mechanical work and on the rostro–caudal organization of muscle activation.

Chapter I
General Methods

Preface

This chapter presents an overview of the experimental methodologies used across the different studies included in this thesis. Rather than providing an exhaustive description of each protocol, the aim is to highlight common methodological frameworks and instrumentation. All studies combined biomechanical, kinematic, and electromyographic data to assess neuromuscular function during walking, particularly in the context of aging, physical activity, and neuromodulation.

Experimental Setup

Across all studies, three core types of data were acquired: ground reaction forces (GRF), three-dimensional (3D) kinematic data, and electromyographic (EMG) activity. These data types were essential for characterizing the mechanical and neuromuscular aspects of gait and balance. To capture them, we employed instrumented treadmills for force measurements, motion capture systems for tracking body segment movements, and wireless EMG systems to record muscle activity. The following sections describe the core instrumentation and data processing methods used across the different experiments included in this thesis.

Ground Reaction Force Measurements

Ground reaction forces were recorded using an instrumented treadmill (h/p/Cosmos-Stellar, Germany) with a belt surface of 1.6 × 0.65 m, equipped with four tri-axial force transducers (Arsalis®, Belgium) mounted beneath the frame. This configuration enabled precise measurement of the vertical (F_z), fore-aft (F_y), and medio-lateral (F_x) components of force (Willems & Gosseye, 2013). Force signals were typically sampled at 1000 Hz, amplified, and filtered using a 4-pole Bessel low-pass filter with a -3 dB cut-off at 200 Hz before being digitized via a 16-bit analog-to-digital converter (e.g., National Instruments PCI-MIO-16E-4). The electrical motor of the treadmill was instrumented with an optical angle encoder to measure the speed of the belt (V_{belt}). Both gait and balance evaluations were conducted on this same instrumented treadmill.

Kinematic Measurements

Second, the kinematic movements of the upper and lower limb segments were recorded in most of the experimental conditions. However, Chapter II employed a different methodology. In that study, no Qualisys motion capture system (Gothenburg, Sweden) was used. Instead, kinematics were recorded using the markerless system FreeMoCap (v1.0.25), which tracked segmental landmarks in the sagittal plane from video recordings at 60 Hz. Anatomical points such as the acromial apex, greater trochanter, lateral femoral condyle, lateral malleolus, and fifth metatarsal head were used to estimate segment orientation and joint angles. The data were filtered using a 4th-order Butterworth low-pass filter (cutoff at 7 Hz) and interpolated to 100 points per stride for further analysis. In contrast, for the remaining studies—particularly those described in Chapters III and IV—recordings were conducted using the Qualisys system, comprising 14 Oqus 600+ and 4 Miqus M3 infrared cameras positioned around the treadmill. A total of 20 retro-reflective markers were placed bilaterally on anatomical landmarks, including the neck, shoulders, elbows, wrists, anterior and posterior superior iliac spines, greater trochanters, knees, malleoli, and fifth metatarsal bones, following the Qualisys Sports marker set.

The kinematic data in these studies were sampled at a frequency between 240 and 250 Hz. To ensure precise temporal alignment with the ground reaction force data (sampled at 1000 Hz), the kinematic signals were up sampled to 1000 Hz using spline interpolation. This approach preserved the temporal resolution of the kinetic data and avoided information loss that could result from down sampling.

Electromyographic Recordings

Surface EMG activity was recorded using Trigno Avanti Sensors from the Delsys Trigno Wireless System (Boston, MA, USA) at a sampling frequency of 2000 Hz. A total of 12 lower-limb muscles on the right side of the body were recorded, including both proximal muscles (e.g., erector spinae [ES], tensor fasciae latae [TFL], gluteus medius [Gmed], vastus medialis [VM], vastus lateralis [VL], rectus femoris [RF]) and distal muscles (e.g., tibialis anterior [TA], semitendinosus [ST], biceps femoris [BF], gastrocnemius lateralis [GL], gastrocnemius medialis [GM], soleus [SOL]). Skin preparation followed the SENIAM guidelines, including shaving, abrasion with sandpaper, and cleaning with ether and alcohol. To ensure accurate electrode placement, muscle bellies were located by palpation and electrodes were oriented along the main direction of the muscle fibers (F. P. Kendall et al., 2005; Winter, 2009). EMG signals were synchronized with kinetic and kinematic data via a Delsys Trigger Module. The raw signals were down sampled to 1000 Hz during post-processing to match the sampling frequency of the other datasets. According to the manufacturer, the Trigno Avanti sensors apply an internal analog band-pass filter (20–450 Hz) to the EMG signal prior to digitization, in order to attenuate movement artifacts and high-frequency noise (Delsys, 2023).

Fig. 1.1 *Participant equipped with 21 retro-reflective markers and 16 wireless EMG sensors on the right side, during treadmill walking.*

Data processing

All experiments in this thesis focused on treadmill walking or balance under controlled conditions. A consistent procedure was applied across all chapters to compute mechanical variables. This section describes the standard approach used to calculate these variables, based on a hypothetical subject walking at constant speed on a treadmill with no incline. Air resistance is considered negligible, and it is assumed that the feet do not skid on the treadmill surface (Cavagna et al., 1976).

Under these conditions, the human body is subjected to two external forces: (1) gravitational force (mg), and (2) ground reaction force (GRF). The GRF is decomposed into three orthogonal components: lateral (F_l), fore-aft (F_f), and vertical (F_v). The corresponding accelerations of the center of mass (CoM) are calculated as:

$$a_l = \frac{F_l}{m},$$

$$a_f = \frac{F_f}{m},$$

$$a_v = \frac{F_v - \overline{F_v}}{m},$$

where m is the body mass $\overline{F_v} \approx BW$.

The computed accelerations were integrated over time to obtain variations in CoM velocity along the vertical (V_v), fore-aft (V_f), and lateral (V_l) axes, each with an associated integration constant. In the fore-aft direction, this constant corresponds to the treadmill V_{belt}, reflecting the steady-state walking condition. In contrast, for the vertical and lateral directions, the integration constants were set to zero, under the assumption that the CoM returns to its initial height and lateral position at the end of each stride.

The displacements of the CoM in the forward (S_f), lateral (S_l) and vertical (S_v) directions relative to the treadmill were obtained by numerically integrating the respective velocity components V_f, V_l and V_v. This procedure assumes that, at the beginning and at the end of the steps analysed, the subject is in the same position on the treadmill

Computation of External Mechanical Work and Power

The external mechanical work (W_{ext}) was calculated as the sum of the positive increments in the total mechanical energy CoM, following the approach proposed by Cavagna et al. (1976):

$$W_{ext} = \sum \Delta E_{com}^{+}$$

The total energy of the CoM (E_{com}) was defined as the sum of the potential energy (E_p), the vertical kinetic energy ($E_{k,v}$) and the horizontal kinetic energy ($E_{k,h}$):

$$E_{com} = E_p + E_{k,v} + E_{k,h}$$

Potential energy was computed as:

$$E_p = g * S_v$$

where S_v is the vertical displacement of the CoM and g is the gravitational acceleration, set to 9.81 m s^{-2}.

Vertical kinetic energy was calculated as:

$$E_{k,v} = \frac{1}{2} m\, v_v^2$$

where v_v is the vertical velocity of the CoM

and horizontal kinetic energy was calculated as:

$$E_{k,h} = \frac{1}{2} m\, v_f^2$$

where v_f is the fore-aft velocity of the CoM

Additionally, in Chapter V, the instantaneous external power of the CoM was calculated as the time derivative of total energy:

$$P_{com} = \frac{dE_{com}}{dt}$$

This power signal was used as a time-resolved variable to evaluate the dynamics of energy generation and absorption by the CoM throughout the gait cycle.

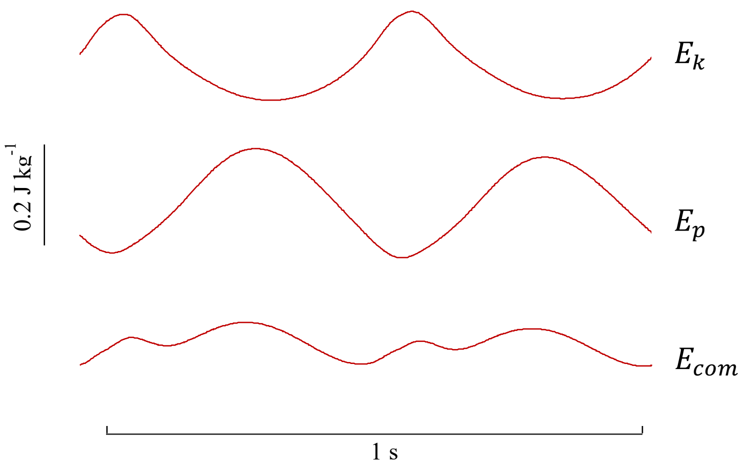

Fig. 1.2 *Typical traces of one representative stride of one participant (1.77 mt and 67.3 kg), illustrating kinetic energy (E_k), potential energy (E_p), and total center of mass energy (E_{com}).*

Statistical Power Computation and Sample Size

Sample sizes were estimated using G*Power v3.1.9.7 in all studies except one. For the study on step-to-step transition strategies in children, data were reused from Schepens et al., (2001), and no a priori power calculation was reported. In all other cases, sample sizes were based on predefined effect sizes and aimed to achieve a power of at least $1-\beta \geq 0.80$. Table 1.1 provides full details on the variables, statistical methods, and software used in each study.

Due to time constraints, an older adult participant voluntarily withdrew from data collection in the tsDCS study. As a result, this participant's data was excluded from the final analysis.

Table 1.1 *Power analysis details for different studies.*

Study + Year of conducted experiments	Software	Statistical test used	Variable/effect size	N Participants for power 1-β ≥ 0.8	Total sample size	Note

Understanding gait alterations: trunk flexion and its effects...(2023)	G*Power v3.1.9.7	One-way ANOVA	F_{back}/F_{front} ratio, Effect size = 1.24, power = 0.85,	7 per condition	20	Based on Dewolf et al.,(2022)
Physical activity in mitigating age-related gait changes (2023)	G*Power v3.1.9.7	One-way ANOVA	Step period/ Effect size = 0.4, power > 0.80	12 per group	59	Based on unpublished correlations; Dewolf et al., (2021)
Acute spinal neuromodulation in older adults (tsDCS) (2023)	G*Power v3.1.9.7	t-test	Walking speed /Effect size = 2.21, power = 0.95	8 per group	40	Based on Awosika et al., (2019)

Note. *Data from the study on step-to-step transition strategies in children were reanalyzed from* (Schepens et al., 2001).

All studies included a sufficient number of participants to restrict type II error, with a statistical power of $1-\beta \geq 0.80$. Nonetheless, most sample sizes remained relatively small. In the study Understanding gait alterations: trunk flexion and its effects, the original sample consisted of 10 male participants; 10 additional female participants were later included following a recommendation from the editor of The Journal of Experimental Biology to improve representativeness. For the studies Physical activity in mitigating age-related gait changes and Acute spinal neuromodulation in older adults (tsDCS), both young and older adults of both sexes were recruited. This balanced design was implemented to ensure that conclusions could be generalized across age and sex, while still controlling for inter-individual variability through careful inclusion criteria.

Inverted Pendulum Model of Walking

The inverted pendulum model provides a fundamental framework for understanding the mechanical behavior of the CoM during walking. As described by Cavagna et al., (1977) this model conceptualizes the stance leg as a rigid inverted pendulum, allowing the body to vault over the supporting limb. During this motion, kinetic energy (E_k) and gravitational potential energy (E_p) fluctuate out of phase, enabling partial energy recovery through passive exchange between these two components.

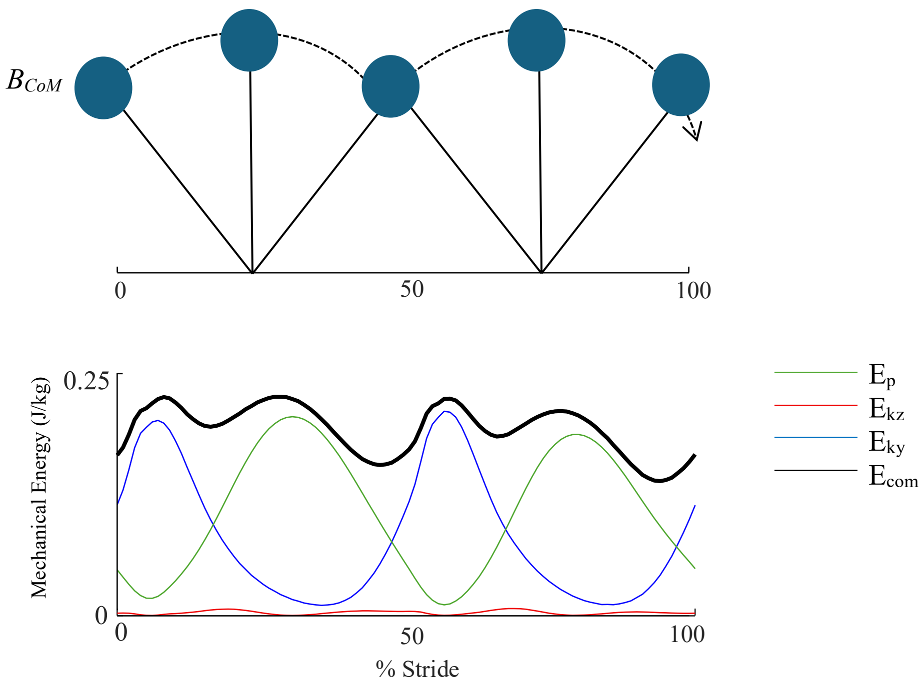

Fig. 1.3 *Schematic illustration of the inverted pendulum model and representative mechanical energy fluctuations during one walking stride of a participant (1.77 m, 67.3 kg). Top: The inverted pendulum representation highlights the arc-like trajectory of the center of mass (BCoM, dotted line) during walking. Bottom: Curves show gravitational potential energy (E_p), vertical kinetic energy (E_{kz}), mediolateral kinetic energy (E_{ky}), and total mechanical energy (E_{com}). External mechanical work (W_{ext}) was computed as the sum of the positive increments in E_{com}.*

External mechanical work (W_{ext}), representing the positive increments in total mechanical energy of the CoM relative to the environment, provides insight into the efficiency of walking. As shown in Fig. 1.3, this energy profile typically reflects an upward-forward displacement of the CoM during mid-stance and a downward motion during double support. This inverted pendulum mechanism is considered mechanically efficient, particularly near preferred walking speeds, and underlies energy-saving strategies in human locomotion (Cavagna et al., 1977; Willems et al., 1995).

Recent review (Peyré-Tartaruga et al., 2021) have emphasized the continued relevance of this model for evaluating gait efficiency. They also highlight that in populations with impaired neuromuscular control, such as older adults or individuals with movement disorders, the behavior of the CoM often deviates from the ideal pendulum mechanism. These deviations include a reduced displacement of the center of mass during walking and a less effective alternation between potential and kinetic energy, which compromises the overall mechanical efficiency of gait.

Step-to-Step Transition Model

To characterize how the CoM is redirected during the double support phase, this thesis builds on the transition strategy framework initially proposed by Chastan et al., (2008). They distinguished two modes of CoM redirection: the active mode, where the downward movement of the CoM is arrested by muscular action—particularly from the plantar flexors—before foot contact (FC); and the passive mode, where this deceleration occurs only upon foot-ground impact, reflecting a more passive reliance on mechanical collision. This framework was subsequently adopted by (Meurisse et al., 2019b) to examine transition control in aging. Methodologically, the detection of FC and toe-off (TO) events used to define the double support phase relies on the approach developed by Meurisse et al., (2016), which identifies these events from the extrema in the center of pressure trajectory. This combination of conceptual and methodological tools allows for a nuanced analysis of gait transition strategies and their modulation across age and functional status (see Fig. 1.4 for an illustration of anticipated vs. non-anticipated transitions, based on the relative timing of CoM vertical velocity minima and foot contact events).

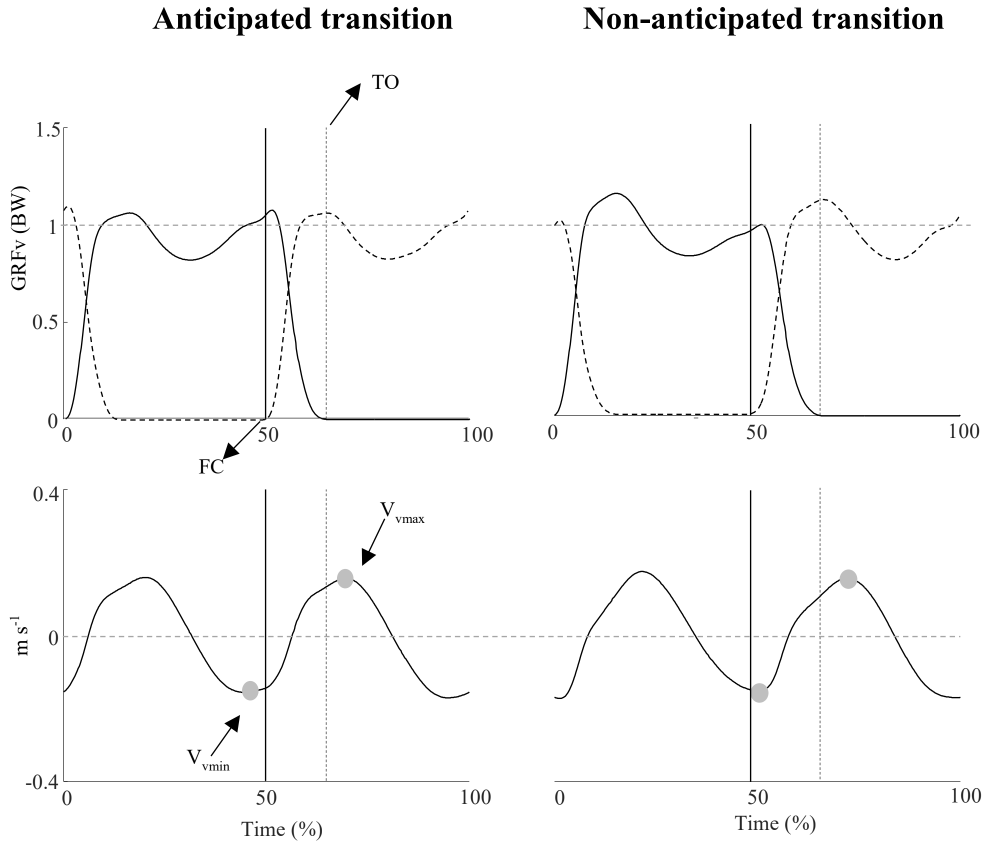

Fig. 1.4 *Illustration of anticipated versus non-anticipated step-to-step transitions based on the timing of the vertical velocity minimum (Vv_{min}) relative to foot contact (FC). In both panels, the upper plots show vertical ground reaction forces (GRFv) from the back leg (dashed line) and front leg (solid line), while the lower plots depict the vertical velocity (Vv) of the CoM. In the anticipated transition (left), Vv_{min} occurs before FC (blue line), indicating early redirection of the CoM. In contrast, in the non-anticipated transition (right), Vv_{min} occurs after FC, suggesting a delayed redirection. The transition phase spans from Vv_{min} to Vv_{max} (gray markers), and the vertical black line denotes TO.*

More recently, Dewolf et al., (2022) expanded this framework by introducing a temporal metric to categorize transitions as anticipated or non-anticipated. This index, which will be used throughout Chapters II, III, IV and

Supplementary I offer a quantifiable approach to assess age- and activity-related differences in gait transition dynamics.

Electromyography Analyses

The raw electromyography (EMG) signals were processed using standardized procedures to ensure comparability across studies. All signals were resampled at 1000 Hz to match the sampling frequency of the kinetic data. A high-pass filter (30 Hz) was first applied to remove movement artefacts, followed by rectification and a zero-lag Butterworth low-pass filter (10 Hz; 4th order). A notch filter at 50 Hz was also implemented to remove frequency-related artefacts. A semi-automatic artefact detection procedure was applied, with gait cycles flagged for review when correlation with the ensemble-averaged EMG envelope fell below 0.6 (Zhvansky et al., 2022). Cycles with evident artefacts (e.g., signal loss or saturation due to electrode detachment) were manually excluded after visual inspection. All accepted gait cycles were time-normalized to 400 points to represent the gait cycle. From the EMG envelopes, several outcome metrics were derived:

- Root mean square (RMS): computed as a measure of overall activation amplitude across the gait cycle. The calculation was performed as:

$$RMS\ (x) = \sqrt{\frac{1}{N}\sum_{i=1}^{N} x_i^2}$$

- Full width at half maximum (FWHM): calculated as the percentage of the gait cycle during which the EMG activity exceeded half of its maximum (Martino et al., 2014; Santuz et al., 2020). Increased FWHM is typically interpreted as broader or less selective activation.
- Centre of activity (CoA): quantified using circular statistics as the angular position of the first trigonometric moment of the EMG distribution across the gait cycle (Batschelet, 1981; Martino et al.,

2014). CoA reflects the relative timing of muscle activation within the gait cycle. The calculation was performed as

$$A = \sum_{i=1}^{400}(\cos(\alpha_i)X\ EMG_i),$$

$$B = \sum_{i=1}^{400}(\sin(\alpha_{i)}X\ EMG_i),$$

$$CoA\ =\ tan^{-1}(B/A)$$

where EMGi represents the EMG amplitude at each phase of the gait cycle. An illustrative schematic of the computation of FWHM and CoA is provided in Fig. 1.5 (adapted from Martino et al., 2014).

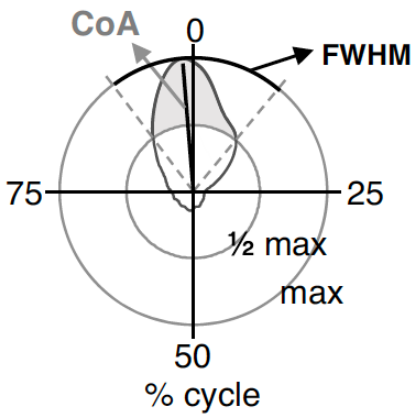

Fig. 1.5. *Schematic representation of FWHM and CoA computation from a representative EMG envelope (adapted from Martino et al., 2014). FWHM corresponds to the portion of the gait cycle in which EMG activity exceeds half of its maximum, while CoA represents the angular position of the first trigonometric moment of the activity distribution.*

- Segmental activation mapping (spinal MN pools): projection of EMG to spinal segments L2–S2 using myotomal charts (Kendall et al., 2005), based on the approximate rostro-caudal location of MN pools innervating different muscles in the human spinal cord. To reconstruct the output pattern of any given spinal segment S_j, all rectified EMG waveforms corresponding to that segment were averaged as:

$$S_j = \frac{\sum_{i=1}^{n_j} k_{ij} \cdot EMG_i}{n_j},$$

where n_j is the number of EMG_i waveforms corresponding to the j^{th} segment, k_{ij} is the weighting coefficient for i^{th} muscle (from Kendall's chart). The assumption implicit in both methods is that the rectified EMG provides an indirect measure of the net firing of MNs of that muscle in the spinal cord. To compute the total motor output for each condition, the motor output patterns across the gait cycle were summed across the lumbar, sacral and all spinal segments. The mean activation of the lumbar (L2 to L5) and sacral (S1 to S2) segments was computed by averaging the motor output patterns for each region.

- The co-activation index (CI) was assessed between the lumbar and sacral segments using the following formula:

$$\mathrm{CI} = \frac{\sum_{j=1}^{400} \{[\mathrm{lumbar}_j + \mathrm{sacral}_j]/2\} \mathrm{x} [\mathrm{Lumbar}_j/\mathrm{sacral}_j]}{400},$$

where lumbar and sacral signals represent the mean activation of each segment across the stride. The CI was then averaged across the full gait cycle, yielding a global estimate of segmental co-activation. High CI values reflect strong simultaneous activation of both segments, whereas low CI values indicate either weak activation of both or dominance of one segment over the other. This index has been previously used to characterize age-related changes in co-activation (Dewolf et al., 2021b).

Depending on the experimental chapter, EMG analyses were conducted at different levels of integration:

- Chapter II: EMG analysis focused on individual lower-limb muscles, specifically the vastus medialis (VM), vastus lateralis (VL), tibialis anterior (TA), medial gastrocnemius (MG), and lateral gastrocnemius (LG). For each muscle, FWHM, CoA, and RMS were quantified.
- Chapters III–IV: In addition to single-muscle analyses, EMG activity was mapped onto the rostro–caudal distribution of spinal motor neuron pools (L2–S2). Segmental motor outputs were extracted for

lumbar (L2–L5) and sacral (S1–S2) regions, and the outcome metrics included FWHM, CoA, RMS, and CI.

This hierarchical EMG approach enabled both muscle-level and spinal-level analyses, allowing the investigation of neuromuscular activation patterns and their modulation by trunk posture, physical activity level, and spinal neuromodulation.

Muscle biomechanical properties

In a standing position, the biomechanical properties of the gastrocnemius medialis (GM) and rectus femoris (RF) were assessed using a handheld, non-invasive myotonometer (MyotonPRO, Myoton AS, Tallinn, Estonia). This device applies a brief mechanical impulse to the muscle belly and records the damped oscillations of the tissue, from which several parameters are computed. For the present study, we focused (on chapter III) on dynamic stiffness and dampening (logarithmic decrement), measured at sites with previously reported high reliability (RF: two-thirds of the distance between the anterior superior iliac spine and the superior pole of the patella; GM: the medial belly at the most reliable site identified by Nguyen et al., 2015).

- Dynamic stiffness (N/m):

$$S = \frac{a_{max} \cdot m}{\Delta l}$$

Where a_{max} is the maximal acceleration of the tissue oscillation, m the effective mass of the testing probe, and Δl the displacement amplitude. This parameter reflects the muscle’s intrinsic resistance to an external force that modifies its shape.

- Logarithmic decrement (dampening):

$$D = \ln\left(\frac{a_1}{a_n + 1}\right)$$

Where $a_1$and $a_n + 1$ represent the amplitudes of two consecutive oscillation peaks. This coefficient quantifies the dissipation of mechanical energy, providing an indirect estimate of the viscoelastic damping properties of the muscle.

These parameters have been previously validated as reliable indices of muscle mechanical behavior in vivo, reflecting intrinsic viscoelastic properties of human skeletal muscle measured under physiological conditions (Garcia-Bernal et al., 2021a; Lettner et al., 2024).

Chapter II

Understanding gait alterations: trunk flexion and its effects on walking neuromechanics

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List of Abbreviations

CoM:	Center of Mass
GRF:	Ground Reaction Force
F_v, F_f, F_l:	Vertical, fore-aft, and lateral components of the GRF
BW:	Body Weight
FC:	Foot Contact
TO:	Toe-Off
DC:	Double Contact (phase)
VM:	Vastus Medialis
VL:	Vastus Lateralis
TA:	Tibialis Anterior
GM:	Medial Gastrocnemius
GL:	Lateral Gastrocnemius
EMG:	Electromyography
ROM:	Range of Motion
V_v, V_f:	Vertical and forward velocity of the center of mass
$V_{v,min}$, $V_{f,min}$:	Minimum vertical and forward velocity of the center of mass
E_k, E_p:	Kinetic Energy, Potential Energy
E_{CoM}:	Total mechanical energy of the center of mass
W_{ext}:	External mechanical work
%R:	Percentage of Recovery (mechanical energy exchange)
FWHM:	Full Width at Half Maximum (EMG signal duration)
CoA:	Center of Activity (timing of EMG activation)
ES:	Effect Size

Abstract

Background: Evolutionary and functional adaptations of morphology and postural tone of the spine and trunk are intrinsically shaped by the field of gravity in which humans move. Gravity also significantly impacts the timing and levels of neuromuscular activation, particularly in foot- support interactions. During step-to-step transitions, the centre of mass velocity must be redirected from downwards to upwards. When walking upright, this redirection is initiated by the trailing leg, propelling the body forward and upward before foot contact of the leading leg, defined as an anticipated transition. In this study, we investigated the neuromechanical adjustments when walking with a bent posture. Twenty adults walked on an instrumented treadmill at 4 km h^{-1} under normal (upright) conditions and with varying degrees of anterior trunk flexion (10, 20, 30 and 40 deg). We recorded lower-limb kinematics, ground reaction forces under each foot, and the electromyography activity of five lower-limb muscles. Our findings indicate that with increasing trunk flexion, there is a lack of these anticipatory step-to-step transitions, and the leading limb performs the redirection after the ground collision. Surprisingly, attenuating distal extensor muscle activity at the end of stance is one of the main impacts of trunk flexion. Our observations may help us to understand the physiological mechanisms and biomechanical regulations underlying our tendency towards an upright posture, as well as possible motor control disturbances in some diseases associated with trunk orientation problems.

Introduction

Upright bipedal walking is one of the most highly automated motor acts that humans perform and distinguishes humans from other mammals. It necessitates several specific adaptations of the locomotor apparatus (Crompton et al., 1998; Spoor et al., 1994) , including the erect posture of the trunk and head. Such an erect posture makes walking gait mechanically efficient because the centre of mass of the body (CoM) vaults over the supporting relatively straight limb like an inverted pendulum (Cavagna & Kaneko, 1977; Willems et al., 1995). This mechanism limits energy expenditure by means of a transduction between the kinetic (E_k) and potential (E_p) energy of the CoM (Cavagna & Kaneko, 1977; A. H. Dewolf et al., 2017; Willems et al., 1995). A major part of the muscular work is used to redirect the CoM from one pendulum arc to the next during walking (Kuo et al., 2005; Ruina et al., 2005). To do so, the propulsion of the trailing limb at the end of stance and the weight acceptance of the leading limb at the beginning of stance are coordinated to avoid collisional loss at foot contact (Dewolf et al., 2022; Dewolf & Willems, 2019; Donelan et al., 2002; Hiebert et al., 1996; Meurisse et al., 2019b). As a result, the lower- limb muscle activation pattern is pulsatile, with muscle activity occurring briefly at specific phases of the cycle to re-excite the intrinsic oscillations of the system when energy is lost (Ivanenko, Poppele, et al., 2004; Lacquaniti et al., 2012).

The head and trunk segments represent up to 65% of the body's mass (Dempster, 1955). Therefore, deviations from the erect posture, even small ones, have inevitable mechanical consequences. For instance, Müller et al., (2017) demonstrated that during walking with a flexed trunk, the CoM is displaced forward and slightly downward, with repercussions on stability (Aminiaghdam et al., 2017, 2018). Also, the associated posterior shift of the hip relative to the CoM leads to a flatter angle at foot contact of the leading leg and a steeper trailing leg angle at toe-off and induces a modification of the two peaks of vertical ground reaction force (GRF) (Aminiaghdam et al., 2016) and a modification of joint moments of the lower limbs (Leteneur et al., 2009), suggesting a modification of the step-to-step transition. These biomechanical modifications lead to alterations in the muscle activation patterns of the lower limb (Alghamdi & Preece, 2020; Grasso et al., 2000), with more crouched postures resulting in greater activation of both proximal and distal muscles, but with a stronger effect on the activation of the thigh and gluteal muscles compared with the shank muscles (Hora et al., 2024). Although these studies provided valuable insights, they did not explore the modification of the step-to-step strategy during the double support phase of walking. Understanding trunk control and its impact on gait is important for clinicians and scientists, as various pathological and ageing-related conditions tend to increase trunk inclination during walking. For example, there is a substantial body of research demonstrating increased trunk flexion during gait with ageing (A. Dewolf, Sylos-Labini, Cappellini, Ivanenko, et al., 2021; A. H. Dewolf, Meurisse, et al., 2019) and with pathologies such as Parkinson's disease (Pongmala et al., 2022) or knee osteoarthritis (Preece & Alghamdi, 2021). In older adults, there is a reduction in mechanical power generated by the plantar flexor muscles during the push-off phase of walking (Delabastita et al., 2021; Winter, Patia, et al., 1990), an unanticipated step-to-step transition (the redirection of the CoM occurring after foot contact) (Dewolf et al., 2022; Meurisse et al., 2019b), and a widening of distal muscle activations (Dewolf et al., 2021a, 2021b)

While the inter-relationship between posture and locomotion is well recognized, the isolated effects of trunk flexion on the neuromuscular control of gait are not well understood. In particular, it is not clear whether the trunk flexion and the associated modification of lower- limb kinematics affect the step-to-step transition of walking and, with it, the activation pattern of lower limb muscles observed in able- bodied human gait. To answer this question, we studied the effects of small (from 10 deg) and large (up to 40 deg) changes in upper body inclination in 20 young and healthy participants. The GRFs, the activity of five lower-limb muscles and the joint angles of lower limb segments were recorded. We hypothesized that trunk flexion would result in a more flexed limb posture during stance, a greater peak of force by the leading limb (and a smaller peak under the trailing one), and an anticipated activation of the distal extensors, consistent with the compensations observed in older adults.

Materials and Methods

Participants

Twenty healthy young adults (10 males and 10 females, age: 29.4±4.6 years, mass: 69.3±11.5 kg, height: 1.69±0.11 m, means±s.d.) volunteered to participate in this study. The sample size was estimated a priori based on the difference of 75% of the rate of Fback/Ffront ($\eta 2=1.51$) (Dewolf et al., 2022). Considering a power analysis of 0.85 ($1-\beta$) and one-way ANOVA with an alpha level of 0.05, 7 participants would be sufficient. The software used for the sample size was G-Power (v.3.1.9.7). All participants had no history of gait pathologies, neurological disease or orthopedic problems that would affect how they walked. Informed consent was obtained from all participants, and the procedures used for this study were approved by the ethics committee of Finis Terrae University, Chile (ref. no. 22-114). The study followed the guidelines of the Declaration of Helsinki. Preliminary results have already been published in a conference paper (Nunez-Lisboa & Dewolf, 2023).

Experimental procedure and data collection

Participants were asked to walk wearing their shoes on an instrumented treadmill at 4 km h^{-1}. The fixed speed imposed in our study allows us to isolate the effect of trunk inclination, as previous research has shown that preferred walking speed can be influenced by trunk inclination (Aminiaghdam et al., 2018). The selected walking speed was reported as the comfortable and economical average walking speed based on the net energy consumption (Cavagna & Kaneko, 1977). Five different gait conditions were tested. Participants were first asked to walk normally, with a natural trunk orientation (called 'normal walking'). Then, they were asked to walk with an anterior trunk flexion of 10, 20, 30 and 40 deg relative to the vertical. The motion capture system FreeMoCap v.1.0.25 (https: /freemocap. org/) was used to track the estimated vertical and anteroposterior coordinates of the acromial apex, greater trochanter, lateral femoral condyle, lateral malleolus and fifth metatarsophalangeal joint at 60 Hz. The participants were filmed in the sagittal plane, and the trunk flexion, defined as the orientation of the line crossing the acromial apex and greater trochanter relative to the vertical axis, during the different gait conditions was controlled by a mobile app (OnForm, v.2.03.0). Before data collection began, participants took several trials with oral feedback from the experimenter until they adopted the correct position. Participants were instructed to bend the hips to achieve the target trunk flexion instead of rounding at the lower or upper back, following the methodology described by Kluger et al., (2014). No instruction was given regarding arm movements. The participants walked then for at least 1 min per condition, with 2 min of rest between each trial to avoid muscular fatigue. Data were recorded for 10 s once steady walking was reached and, on average, 7.3±0.5 (mean±s.d.) strides were analyzed for each participant.

The treadmill (Stellar, h/p/cosmos, Nußdorf, Bayern, Germany; belt surface: 1.6×0.65 m; mass: ~240 kg) was instrumented with four force transducers (Arsalis®, Genappe, Belgium). As the transducers were placed under the body of the treadmill, the force transducers measured the three components of GRF exerted by the treadmill belt under the foot (Willems & Gosseye, 2013): Fv, Ff and Fl are, respectively, the vertical, fore aft and lateral component of the GRF. Data were sampled at a frequency of 500 Hz. Two algorithms were used to reconstruct the three components of the force under the left and right foot (Bastien et al., 2019; Meurisse et al., 2016a).

Electromyographic (EMG) activities from five muscles of the right lower-limb were recorded at 2 kHz using a Delsys Trigno Wireless System (Boston, MA, USA): vastus medialis (VM), vastus lateralis (VL), tibialis anterior (TA), medial gastrocnemius (MG) and lateral gastrocnemius (LG). We selected these muscles to have a good comparison of activity between proximal and distal muscles of the lower limbs. The EMG activity of other relevant muscles, such as the biceps femoris, rectus femoris and gluteus maximus, has previously been reported in a similar task (Grasso et al., 2000). Furthermore, our focus was specifically on step-to-step transitions, where extensor muscles have been shown to significantly contribute to CoM propulsion and where the recorded muscles are particularly active (A. H. Dewolf, Ivanenko, et al., 2019a; Ivanenko et al., 2008). The location of the electrode was based on suggestions from SENIAM (the European project of surface EMG, seniam.org). The signal quality was verified visually by checking the EMG signals during voluntary contractions before walking. Acquisition of the EMG and GRF were synchronized with a trigger module from Delsys.

Data analysis

<u>Division of the stride</u>

The foot contact (FC) and toe-off (TO) events were estimated from the displacement of the centre of pressure on the belt (Meurisse et al., 2016a). A stride was defined as two successive right FCs. Stance phases were measured as the time between TO and FC of the same leg. The double contact phase (DC) started from the front leg FC to the back leg TO.

<u>Kinematics</u>

The vertical and antero-posterior coordinates of the head of the fifth metatarsal, the external malleolus, the lateral condyle of the knee, the greater trochanter and the apex of the shoulder of the right side of the body were recorded. The coordinates were filtered with a 4th order Butterworth filter, with a 7 Hz cut-off frequency. The angle of each segment (trunk, thigh, shank and foot) relative to the vertical (elevation angle) was computed in the sagittal plane. From these angles, the hip, knee and ankle joint angles were calculated. The hip, knee and ankle joint angles during standing were subtracted from those of the joint measured during walking tasks. Each stride of each participant was interpolated over 100 points. For the three joint angles and the trunk elevation angle, the range of motion (ROM) and the mean values were measured. Joint angles were measured relative to the subject's standing position, with 0 deg indicating this posture.

<u>Kinetics</u>

The fore–aft and vertical velocity of the CoM were determined from the fore–aft and vertical components of the GRF using the procedure described in detail in (A. H. Dewolf et al., 2017). In short, the fore–aft acceleration of the CoM was calculated as $a_f = F_f / m$, where m is the participant's body mass. The vertical acceleration of the CoM was calculated as $a_v = (F_v - m\text{g})/m$, where g is the acceleration due to gravity. The vertical (V_v) and the forward velocity (V_f) of the CoM were calculated by time-integration of a_v and a_f, respectively, plus an integration constant, which was computed so that the average velocity over a stride was equal to zero. The vertical and forward displacements of the CoM (S_v and S_f, respectively) were then computed by time-integration of V_v and V_f.

The time of occurrence of the minimum of forward ($V_{f,min}$) and vertical velocities of the CoM ($V_{v,min}$) at the beginning of the double contact phase were detected. The step-to-step transition strategy was defined as follows: the transition was considered as anticipated when $V_{v,min}$ occurred before FC and unanticipated when $V_{v,min}$ occurred after FC. Also, the peaks of vertical force under the front leg (F_{front}) and the back leg (F_{back}) were measured during the DC. In addition, the negative ($F_{f,brake}$) and positive ($F_{f,propulsion}$) peak of the left and right fore–aft force were measured.

The energy of the CoM (E_{CoM}) was computed as the algebraic sum at each instant of its gravitational potential energy (E_p=mgS_v) and its kinetic energy [E_k=½$m(V_f +V_v)^2$]. The work done to move the CoM relative to the surroundings (W_{ext}) was then computed as the sum of the positive increments of E_{CoM} (Cavagna et al., 1976, 1977). The energy transduction between E_p and E_k was estimated from the relative amount of energy saved over a step (%R):

$$\%R = \frac{W_k + W_p - W_{ext}}{W_k + W_p} * 100$$

EMG

The raw EMG signals were high-pass filtered (30 Hz), rectified and low-pass filtered with a zero-lag 3rd order Butterworth filter (10 Hz). The time scale was normalized by interpolating individual gait cycles over 400 points. For each condition and each stride, the full width at half maximum (FWHM) was calculated as the period during which the EMG activity exceeded half of its maximum (Martino et al., 2014). Increased FWHM may indicate altered motor control and joint instability (Martino et al., 2014). Each EMG waveform's centre of activity (CoA) was calculated as the vector's angle that points to the CoM of the circular distribution (Martino et al., 2014). Also, the mean activation was calculated.

Statistics

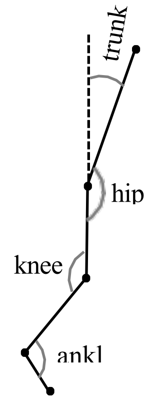
For each dependent variable, the normality of residuals was visually assessed using Q–Q plots. A mixed-effects model with a post hoc Fisher LSD test was performed to evaluate differences between normal and the 10, 20, 30 and 40 deg conditions. Both numerator and denominator degrees of freedom (d.f.) were reported; the numerator d.f. reflects the number of comparisons, and the denominator d.f. adjusts for the sample size and variability. In cases where sphericity was violated, the Greenhouse–Geisser correction was applied to adjust both d.f. (Wright & Wolfinger, 1997). Furthermore, the Friedman test with post hoc Dunn's test was used for dependent variables that did not meet the normality assumption. Cohen's d-effect size (ES) was calculated for each comparison, interpreted as 0.2 (small), 0.5 (medium) and 0.8 or greater (large). All statistical analyses were conducted using GraphPad Prism 9 software (GraphPad Software, San Diego, CA, USA) with an alpha level set at 0.05.

Results

Fig. 2.1A and B illustrates the digital joint identification at the sagittal plane, the typical curves of the vertical and fore–aft forces of the back and front leg, and raw EMG data during one stride in all conditions. Fig. 2.2 demonstrates the average comparison of walking conditions with trunk flexion. The anterior trunk flexion during walking differed between conditions ($F_{2.857,50.72}$=252.1, P<0.0001; Fig. 2.2). Specifically, compared with normal walking, the degrees of anterior trunk flexion were higher at 10, 20, 30 and 40 deg (post hoc: $p < 0.0001$, ES=4.31; $p < 0.0001$, ES=7.61; $p < 0.0001$, ES=6.44; $p < 0.0001$, ES=8.55, respectively), and the average trunk elevation angle was close to (but smaller than) the angle requested to the participants (normal=−1.55±1.92 deg; 10 deg=7.5 ± 2.2 deg; 20 deg=18.3±3.1 deg; 30 deg=24.5±5.3 deg; 40 deg=33.3 ± 5.3 deg). The stride duration was affected by the trunk flexion ($F_{1.644,31.24}$=17.40, $p < 0.0001$; Fig. 2.2). Specifically, compared with normal walking, the stride duration was lower at 10, 20, 30 and 40 deg (post hoc: p=0.045, ES=0.26; p=0.005, ES=0.30; p=0.001, ES=0.83; p<0.0001, ES=1.25, respectively). However, no difference was observed in the stance duration ($F_{2.439,45.72}$=1.865, P=0.158; Fig. 2.2).

Specifically, compared with normal walking, the stance duration did not change at 10, 20, 30 and 40 deg (post hoc: p=0.938, ES=0.14; p=0.778, ES=0.12; p=0.734, ES=0.32; p=0.247, ES=0.59, respectively.

A



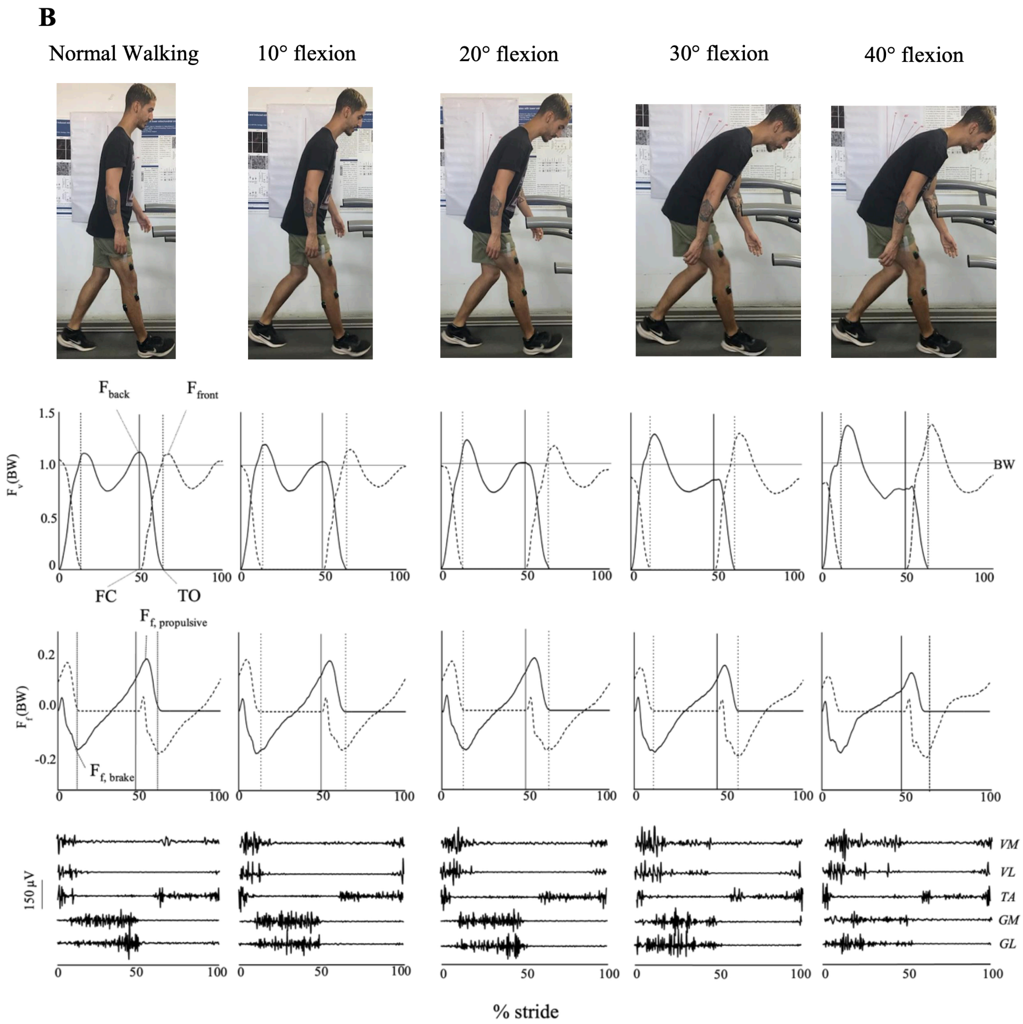


Fig. 2.1. *Digital joint identification at the sagittal plane, typical traces of one representative stride in each condition showing the ground reaction force, raw signals of muscle activity, and spatial-temporal parameters of walking. (**A**) The picture illustrates the digital joint identification at the sagittal plane using Freemocap software, with the vertical reference line and joint markers (shoulder, hip, knee, and ankle) highlighted. (**B**) The pictures illustrate one participant in each condition (normal walking, 10° flexion, 20° flexion, 30° flexion, and 40° flexion) during the double-contact phase of walking. The curves below represent the vertical and forward ground reaction force normalized by body weight (F_v and F_f, respectively), acting upon each leg separately (continuous line: front leg; dotted line: back leg). The peak of vertical force under the back leg (F_{back}) and the front leg (F_{front}) and the peak of negative ($F_{f, brake}$) and positive ($F_{f, propulsive}$) fore-aft force. The vertical continuous and dotted lines indicate the foot contact (FC) and toe-off (TO), respectively.(C) Typical raw*

EMG signals of the vastus medialis and lateralis (VM and VL), anterior tibialis (TA), and medial and lateral gastrocnemius (GM and GL) at each trunk inclination.

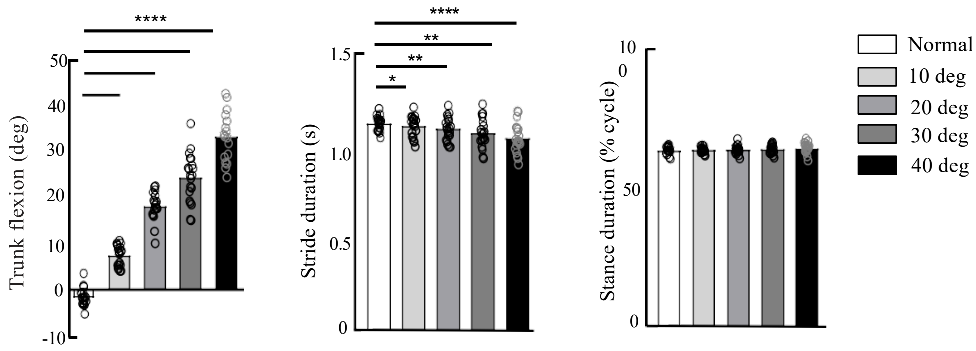
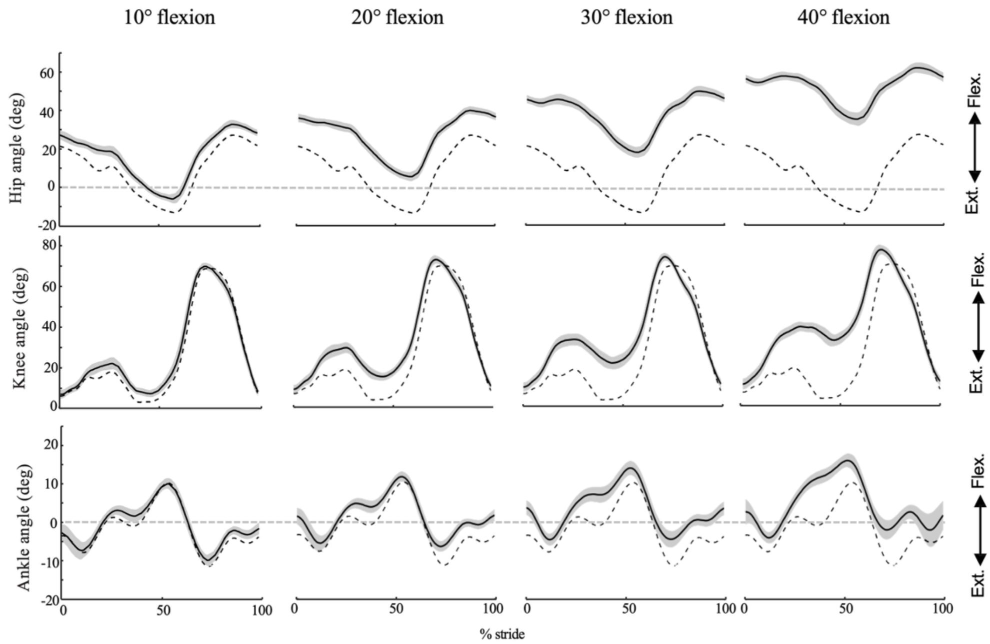


Fig. 2.2. *Spatio-temporal parameters of walking. Left: anterior trunk flexion during walking. Middle: stride duration relative to increasing trunk flexion. Right: stance duration as a percentage of cycle duration relative to increasing trunk flexion. The bars represent the mean, and the circles represent the mean values for each condition participant. The asterisks denote post hoc comparisons between the normal condition and the other conditions (*P<0.05, **P<0.01, ***P<0.0001).*

Fig. 2.3A illustrates the average waveform of the hip, knee and ankle joint angles during normal and flexed conditions over a stride. Notably, the joint angle's time course changed significantly as trunk flexion increased during the stride. Trunk inclination significantly affected the average angles of all lower-limb joints (hip: $F_{1.200,118.8}=4404$, $p<0.0001$; knee: $F_{1.058,104.8}=124.6$, $p<0.0001$; ankle: $F_{1.588,157.2}=404.3$, $p<0.0001$) (Fig. 2.3B). Furthermore, trunk inclination had a significant impact on the mean ROM of the hip, knee and ankle joint (hip: $F_{3.180,56.44}=165.7$, $p<0.0001$; knee: $F_{2.816,49.99}=28.60$, $p<0.0001$; ankle: $F_{2.776,49.27}=15.48$, $p<0.0001$) (Fig. 2.3B). Compared with the normal condition, the average joint angle of the hip was higher at 10, 20, 30 and 40 deg of trunk flexion (post hoc: $p<0.0001$, ES=2.05; $p<0.0001$, ES=4.17; $p<0.0001$, ES=5.36; $p<0.0001$, ES=7.34, respectively), while the average joint angle of the ankle and knee was higher at 20, 30 and 40 deg of trunk flexion (ankle post hoc: $p=0.204$, ES=0.37; $p<0.0001$, ES=1.14; $p<0.0001$, ES=1.68; $p<0.0001$, ES=2.21, respectively) (knee post hoc:

p=0.080, ES=0.59; p<0.0001, ES=1.75; p<0.0001, ES=2.40; p<0.0001, ES=3.09, respectively).

A



B

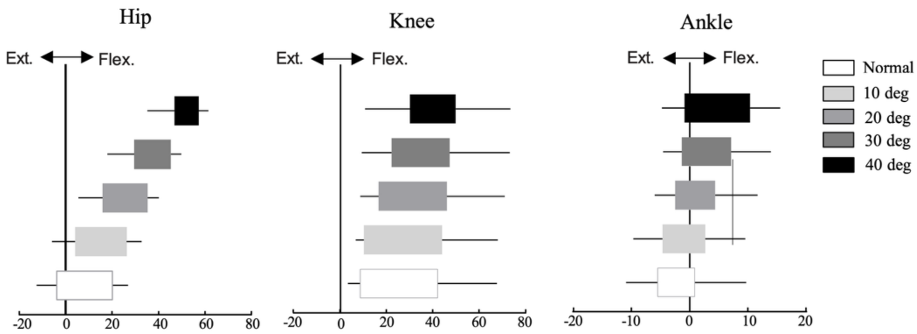
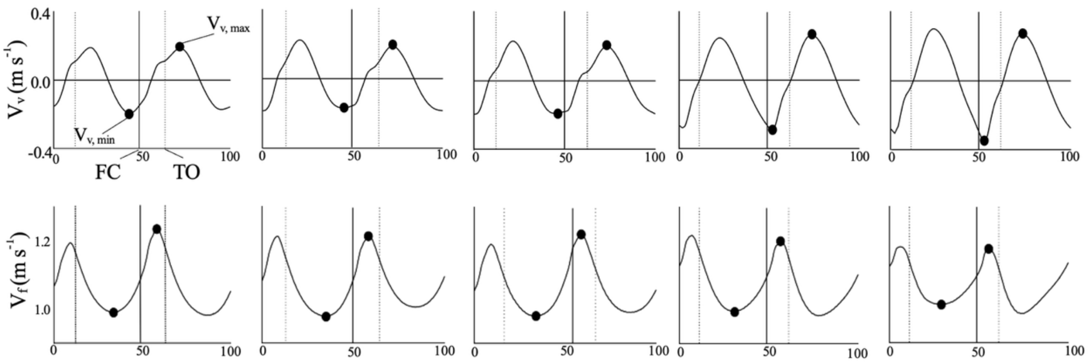


Fig. 2.3. *Joint angles of the lower limbs during walking. (**A**) The joint angle of the hip (top), knee (middle) and ankle (bottom) relative to the percentage of the stride at 4 km h^{-1} in each condition. All the curves of each participant walking at a given trunk inclination condition were first averaged (mean curve). The dashed black curve represents the average joint angle of normal walking. The solid black curve and the grey zone represent the average joint angle and the standard deviation, respectively, with 10, 20, 30 and 40 deg of trunk inclination. (**B**)*

The average range of motion of the hip, knee and foot over one stride at 4 km h^{-1}. Each participant's strides when walking at a given trunk inclination condition were first averaged. The rectangles represent the grand mean of each condition at the hip, knee and ankle joints. Error bars represent the standard deviation. In A and B, an angle of 0 deg corresponds to a standing position; positive values represent a more flexed joint than during standing.

Fig. 2.4A presents typical traces of V_v (top) and V_f (bottom) during one stride in all walking conditions. The black circles represent the minimum ($V_{v,min}$) and maximum ($V_{v,max}$) of V_v, which define the step-to-step transition. The black circles also indicate the minimum ($V_{f,min}$) and maximum ($V_{f,max}$) of V_f. With increasing trunk flexion angle, $V_{v,min}$ occurred later in the stride ($\chi 2$=68.98, d.f.=4, p<0.0001). This indicates that the step-to-step transition was anticipated during normal walking and walking with a 10 deg flexion ($V_{v,min}$ occurred before FC) and unanticipated at higher trunk flexion angles ($V_{v,min}$ occurred after FC) (post hoc: 20 deg: p<0.0001, ES=1.64; 30 deg: p<0.0001, ES=2.47; 40 deg: p<0.0001, ES=2.88) (Fig. 2.4B). In a parallel observation, as the trunk flexion angle increased, $V_{f,min}$ was noted to occur before heel strike ($F_{1.534,28.77}$=3.139, p=0.0703), coinciding with a monotonic change in the peak of vertical and fore– aft forces exerted by the back leg and by the front leg (Fig. 2.4C). With increasing trunk flexion, F_{back} decreased ($F_{2.165,41.13}$=84.31, p<0.0001). Compared with normal walking, the mean F_{back} decreased at 10, 20, 30 and 40 deg (post hoc: p<0.0001, ES=1.03; p<0.0001, ES=1.85; p<0.0001, ES=2.54; p<0.0001, ES=2.72, respectively). In contrast, F_{front} increased with increasing of trunk flexion ($F_{2.021,38.40}$=16.04, p<0.0001). The mean F_{front} increased at 20, 30 and 40 deg compared with normal walking (post hoc: p=0.002, ES=0.67; p<0.0001, ES=1.09; p=0.0004, ES=0.99, respectively) (Fig. 2.4C). Additionally, $F_{f,propulsive}$ decreased with increasing trunk flexion ($F_{2.351,44.08}$=32.64, p<0.0001). Specifically, compared with normal walking, the mean $F_{f,propulsive}$ decreased at 10, 20, 30 and 40 deg (post hoc: p=0.037, ES=0.42; p=0.001, ES=0.95; p<0.0001, ES=1.61; p<0.0001, ES=2.10, respectively). Conversely, $F_{f,brake}$ was unaffected by trunk inclination ($F_{2.023,37.94}$=0.4022, p=0.674). Specifically, there were no significant changes in mean $F_{f,brake}$ at 10, 20, 30 and 40 deg (post hoc: p=0.192, ES=0.21; p=0.278, ES=0.22; p=0.787, ES=0.07; p=0.609, ES=0.13, respectively) (Fig. 2.4C).

A



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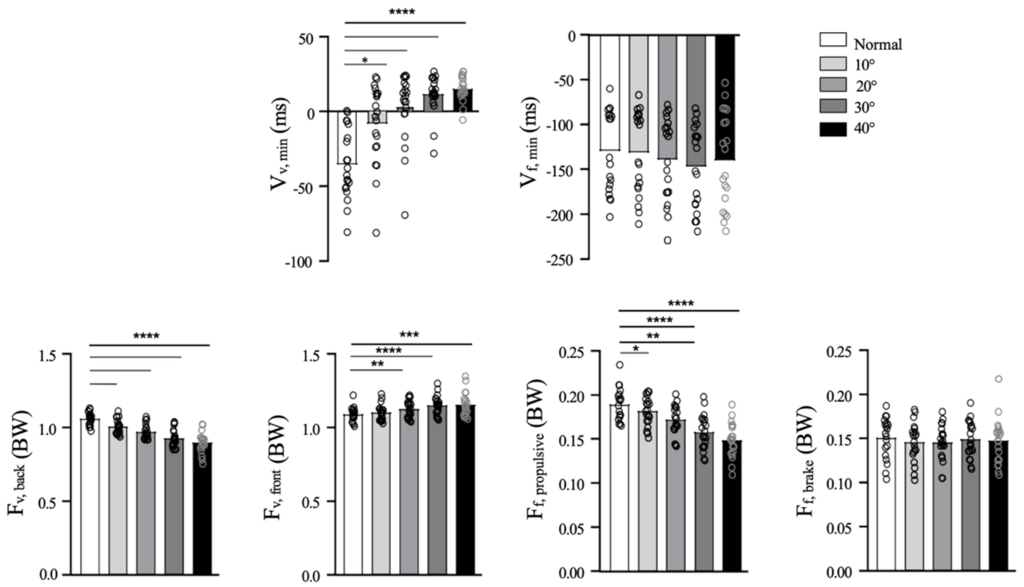
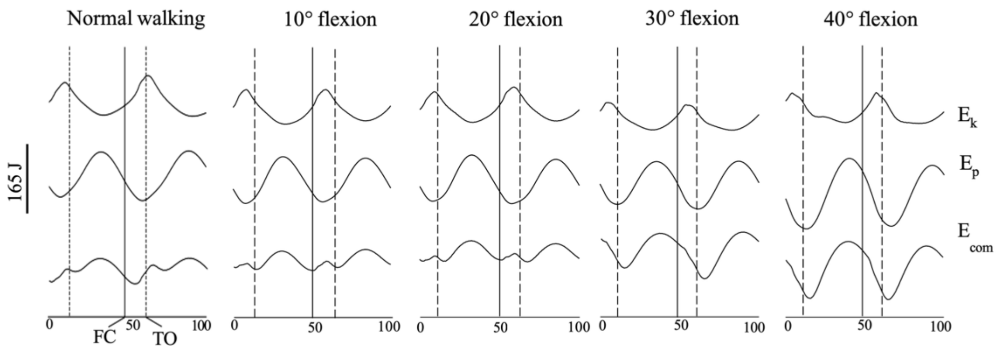


Fig. 2.4. *Step-to-step transition strategy. (**A**) The top and bottom graphs represent the vertical (Vv) and forward (Vf) velocity of the centre of mass (CoM), respectively, at each trunk inclination. The black circles indicate the minimal ($V_{v,min}$ and $V_{f,min}$) and maximal ($V_{v,max}$ and $V_{f,max}$) velocity of the CoM. (**B**) Occurrence of the $V_{v,min}$ (left) and $V_{f,min}$ (right) of the CoM relative to FC with increasing trunk flexion during walking. (**C**) Vertical and forward peak forces normalized to body weight for the back leg (F_{back} and $F_{f,propulsive}$) and front leg (F_{front} and $F_{f,brake}$) as a function of trunk flexion. Other indications as in Fig. 2.2 *P<0.05, **P<0.01, ***P<0.001; ****P<0.0001.*

Fig. 2.5 demonstrates typical traces of the pendulum mechanism variables. Specifically, Fig. 2.5A illustrates a typical E_k, E_p and E_{CoM} trace during one stride under each condition. The CoM work (W_{ext}) significantly differed between conditions ($F_{1.292,24.55}$=7.572, *p*=0.007, Fig. 2.5B). Compared with normal walking, W_{ext} was lower at 10 and 20 deg (post hoc: $_p$=0.0004, ES=0.91; *p*=0.006, ES=0.84, respectively) (Fig. 2.5B). Additionally, W_{ext} did not change at 30 and 40 deg (*p*=0.415, ES=0.26; *p*=0.096, ES=0.58, respectively). The modification of W_{ext} may partly be explained by the change of pendulum-like energy exchange (Fig. 2.5B). Indeed, the inclination of the trunk affects the %R ($F_{1.387,26.35}$=11.82, *p*=0.0008). As compared with normal walking, the %R was higher at 10 and 20 deg (post hoc: *p*=0.0004, ES=0.87; P=0.015, ES=0.81, respectively), with no changes at 30 deg (*p*=0.405, ES=0.39), and was significantly lower at 40 deg (post hoc: *p*=0.025, ES=0.84).

A



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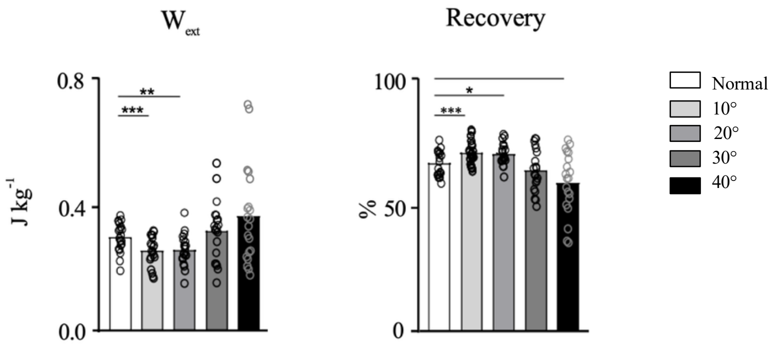

Fig. 2.5. *Pendulum mechanism of walking. (**A**) Typical traces of one representative stride of one participant (same as in Fig. 1) in each condition illustrating kinetic (E_k), potential (E_p) and centre of mass energy (E_{CoM}) with different trunk flexion. Other indications as in Fig. 2.2 (**B**) Positive external work to move the CoM relative to the surroundings over one stride (W_{ext}) and average recovery (%R) during the stride with increasing trunk flexion. Other indications as in Fig. 2. *P<0.05, **P<0.01, ***P<0.001.*

Fig. 2.6 presents the average (across strides and participants) rectified muscle activity of lower limb muscles (VM, VL, TA, GM and GL) during one stride in each condition. The angle of trunk flexion significantly influenced the mean activation of most muscles (VM: $F_{1.563,19.15}$=12.99, p=0.0006; VL: $F_{3.018,46.03}$=4.153, p=0.010; TA: $F_{1.530,16.06}$=8.180, p=0.0058; GL: $F_{1.619,23.87}$=5.733, p=0.013), except for GM, which was not significantly affected ($F_{1.787,22.79}$=1.583, p=0.227; Fig. 2.6B, top). Specifically, compared with normal walking, the mean activation of GM did not change at 10, 20, 30 and 40 deg (post hoc: p=0.335, ES=0.41; p=0.432, ES=0.33; p=0.954, ES=0.23; p=0.904, ES=0.21, respectively). Furthermore, compared with normal walking, the mean VM muscle activity increased at 10, 30 and 40 deg (post hoc: p=0.035, ES=0.16; p=0.005, ES=0.65; p<0.0001, ES=1.20, respectively). However, VM activity did not change at 20 deg of trunk flexion (post hoc: p=0.263, ES=0.24). Additionally, the mean VL muscle activity increased at 10 and 40 deg (post hoc: p=0.020, ES=0.31; p=0.031, ES=0.65, respectively) and did not change at 20 and 30 deg (post hoc: p=0.118, ES=0.40; p=0.189, ES=0.51, respectively). The mean TA activity was higher at 40 deg (post hoc: p=0.0174) and did not change at 10, 20 and 30 deg (post hoc: p=0.686, ES=0.05; p=0.352, ES=0.15; p=0.091, ES=0.38, respectively). Lastly, the mean activation of GL was higher at 30 and 40 deg (post hoc: p=0.039, ES=0.51; P=0.008, ES=0.65, respectively), and did not change at 10 and 20 deg of trunk flexion (post hoc: p=0.353, ES=0.10; p=0.114, ES=0.35).

In most muscles, the duration of muscle activation also varied with trunk flexion (Fig. 2.6B, middle). The trunk inclination affected the FWHM of VM, VL, GM and GL (VM: $F_{2.510,30.12}$=12.46, p<0.0001; VL: $F_{2.288,37.76}$=4.604, p=0.0129; GM: $F_{3.108,52.06}$=3.217, p=0.028; GL: $F_{2.991,56.07}$=5.515, p=0.002). In contrast, the FWHM of TA was not significantly affected by trunk angle ($F_{2.170,40.69}$=0.2317, p=0.811). Specifically, compared with normal walking, the mean FWHM of TA did not change at 10, 20, 30 and 40 deg of trunk flexion (post hoc: p=0.976, ES=0.0; p=0.993, ES=0.0; p=0.636, ES=0.11; p=0.670, ES=0.11, respectively). Furthermore, compared with normal walking, the FWHM of VM was higher at 10, 20, 30 and 40 deg of trunk flexion (post hoc: p=0.005, ES=0.47; p=0.004, ES=0.93; p=0.0007, ES=1.43; p=0.0001, ES=1.57, respectively). The FWHM of VL was higher at 40 deg of trunk flexion (post hoc: p=0.021, ES=0.89), and did not change at 10, 20 and 30 deg of trunk flexion (post hoc: p=0.656, ES=0.12; p=0.928, ES=0.03; p=0.361, ES=0.27, respectively). The FWHM of GM was higher at 20, 30 and 40 deg (post hoc: p=0.015, ES=0.95; p=0.017, ES=0.84; p=0.014, ES=0.81, respectively), and did not change at 10 deg (post hoc: p=0.147, ES=0.47). The FWHM of GL was higher at 20, 30 and 40 deg of trunk flexion (post hoc: p=0.016, ES=0.47; p=0.0004, ES=0.76; p=0.045, ES=0.44, respectively), and did not change at 10 deg of trunk flexion (post hoc: p=0.071, ES=0.23).

The trunk flexion also influenced the timing of activation of extensor muscles (VM: $F_{2.800,30.80}$=16.09, p<0.0001; VL: $F_{2.158,38.31}$=12.95, p<0.0001; GM: $\chi2$=40.6, d.f.=4, p<0.0001; GL: $F_{2.272,30.67}$=19.07, p<0.0001) but not of flexor muscles (TA: $F_{2.881,54.01}$=0.3533, p=0.778) (Fig. 2.6B, bottom). Compared with normal walking, the activation of VM occurred earlier at 30 and 40 deg of flexion (post hoc: p=0.0003, ES=1.19; p=0.0008, ES=1.55) and did not change at 10 and 20 deg of flexion (post hoc: p=392, ES=0.11; p=0.889, ES=0.03, respectively). Furthermore, the activation of VL occurred later at 30 and 40 deg of flexion (post hoc: p=0.0006, ES=1.29; p=0.0031, ES=1.41, respectively) and did not change at 10 and 20 deg of trunk flexion (post hoc: p=0.392, ES=0.05; p=0.344, ES=0.67, respectively). For distal extensors, the changes were much larger: compared with normal walking, the CoA of GM and GL occurred earlier at all trunk flexion angles (GM: 10 deg: p=0.021, ES=65; 20 deg: p=0.0001, ES=1.15; 30 deg: p<0.0001, ES=1,30; 40 deg: p<0.0001, ES=1.69; GL: 10 deg: p=0.0002, ES=0.88; 20 deg: p=0.0002, ES=1.29; 30 deg: p=0.0003, ES=2.25; 40 deg: p<0.0001, ES=2.20).

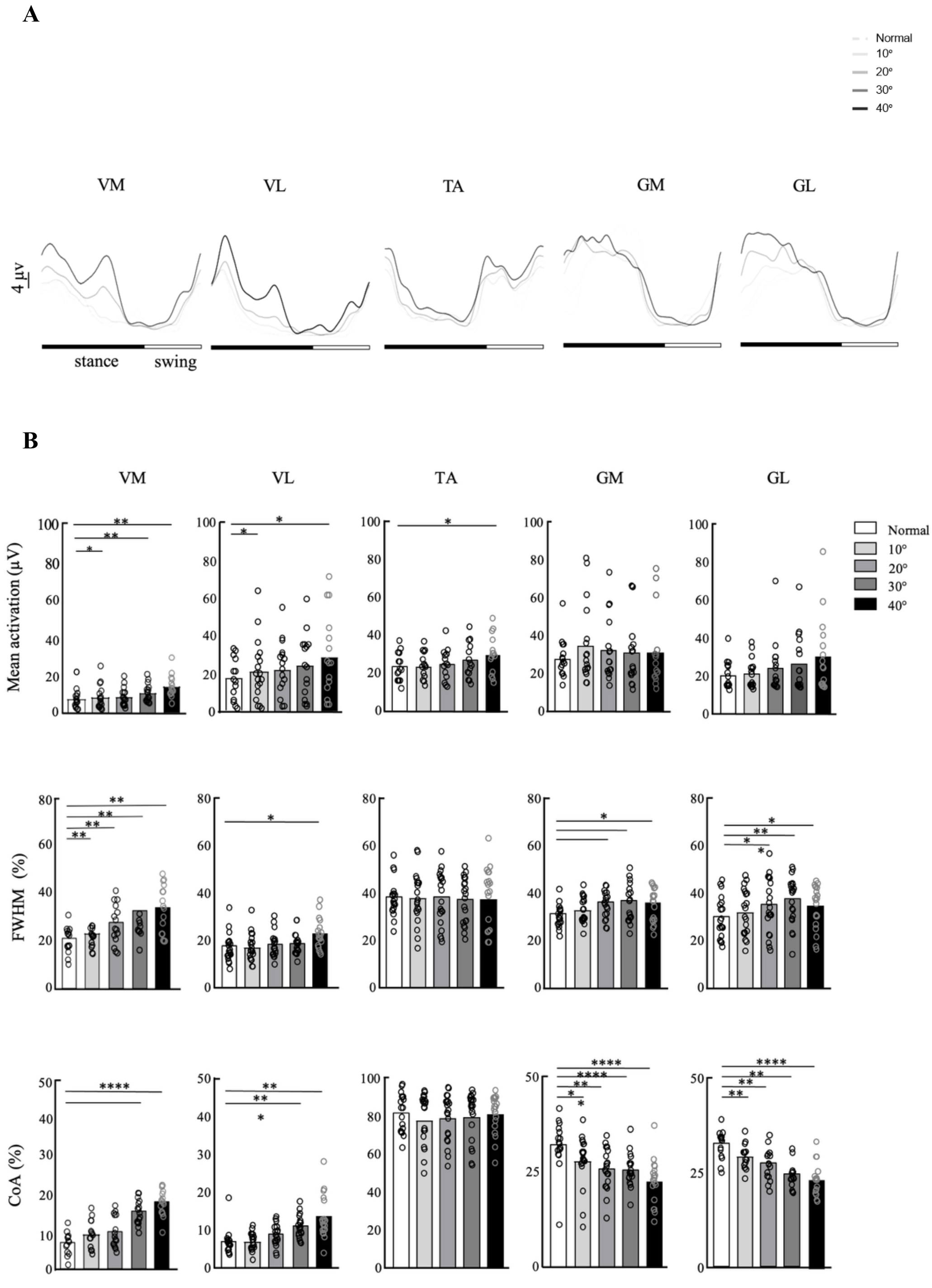
A
Normal
10º
20º
30º
40º
VM
VL
TA
GM
GL
4 µv
stance
swing
B
VM
VL
TA
GM
GL
Normal
10º
20º
30º
40º
Mean activation (µV)
FWHM (%)
CoA (%)

*Fig. 2.6 Muscular activity during walking. (**A**) Typical traces of the filtered and rectified EMG amplitude of one representative stride of one participant (same as in Fig.2.1) in each condition. The black horizontal line represents the duration of the stance phase, whereas the white line represents the duration of the swing phase. The y-axis indicates the amplitude of the EMG activity, with a vertical bar representing 4 μV as a reference. (**B**) Mean activation (top), full width at half maximum (FWHM, middle) and centre of activity (CoA, bottom) of the six lower-limb muscles as a function of trunk flexion. VM, vastus medialis; VL, vastus lateralis; TA, tibialis anterior; GM, medial gastrocnemius; GL, lateral gastrocnemius. Other indications as in Fig. 2. *p<0.05, **p<0.01, ***p<0.001; ****p<0.0001.*

Fig. 2.7A presents the mean activation of the extensor muscles (VM, GM and GL) during the initial and final 20% of the stance phase. The initial 20% of the stance phase was influenced by trunk flexion in most muscles (VM; $F_{2.646,28.44}$=18.70, p<0.0001; GM: $F_{2.586,34.27}$=4.714, p=0.009; GL: $F_{2.900,52.92}$=4.928, p=0.004). Compared with the normal condition, the mean activation of VM during this phase was higher at 10, 30 and 40 deg of trunk flexion (*post hoc*: p=0.040, ES=1.21; p=0.0006, ES=4.15; p=0.0003, ES=5.52, respectively) and did not change at 10 deg of trunk flexion (*post hoc*: p=0.096, ES=0.99). In contrast, VL was not significantly affected ($F_{1.878,26.30}$=1.150, p=0.329). Compared with normal walking, VL did not change at 10, 20, 30 and 40 deg of trunk flexion (*post hoc*: p=0.921, ES=0.02; p=0.095, ES=0.32; p=0.301, ES=0.15; p=0.195, ES=0.21, respectively). The mean activation of GM was higher at 20 and 30 deg (*post hoc*: p=0.032, ES=0.86; p=0.016, ES=0.91, respectively) and did not change at 10 and 40 deg of trunk flexion (*post hoc*: p=0.188, ES=0.48; p=0.127, ES=0.73, respectively). The mean activation of GL was higher at 30 and 40 deg (*post hoc*: p=0.047, ES=0.65; p=0.016, ES=0.77, respectively) and did not change at 10 and 20 deg (*post hoc*: p=0.412, ES=0.19; p=0.084, ES=0.54, respectively).

Regarding the final 20% of the stance phase, discernible alterations were observed in VM ($F_{2.630,32.21}$=3.016, p=0.0501). Specifically, VM remained significantly unaltered at 10, 20, 30 and 40 deg of trunk flexion (*post hoc*: p=0.622, ES=0.19; p=0.695, ES=0.17; p=0.114, ES=0.81; p=0.061, ES=0.86, respectively) (Fig. 2.7A). Similarly, VL did not show significant changes ($F_{0.06431,0.9325}$=1.210, p=0.1214). Compared with normal walking, VL did not report significant changes (*post hoc*: p=0.884, ES=0.03; p=0.756, ES=0.06; p=0.142, ES=0.42; p=0.161, ES=0.39, respectively). Furthermore, the mean activation of GM did not present significant changes (χ^2=0.5, d.f.=4, *p*=0.969). Specifically, GM did not change at 10, 20, 30 and 40 deg of trunk flexion (*post hoc*: *p*=0.932, ES=0.06; *p*=0.682, ES=0.01; *p*=0.969, ES=0.14; *p*=0.622, ES=0.14, respectively). In contrast, GL showed significant changes (χ^2: 12.33, d.f.=4, *p*=0.015), increasing at 40 deg of trunk flexion (*post hoc*: *p*=0.017, ES=0.64) but with no change at 10, 20 and 30 deg of trunk flexion (*post hoc*: *p*=0.656, ES=0.14; *p*=0.806, ES=0.08; *p*=0.066, ES=0.55, respectively).

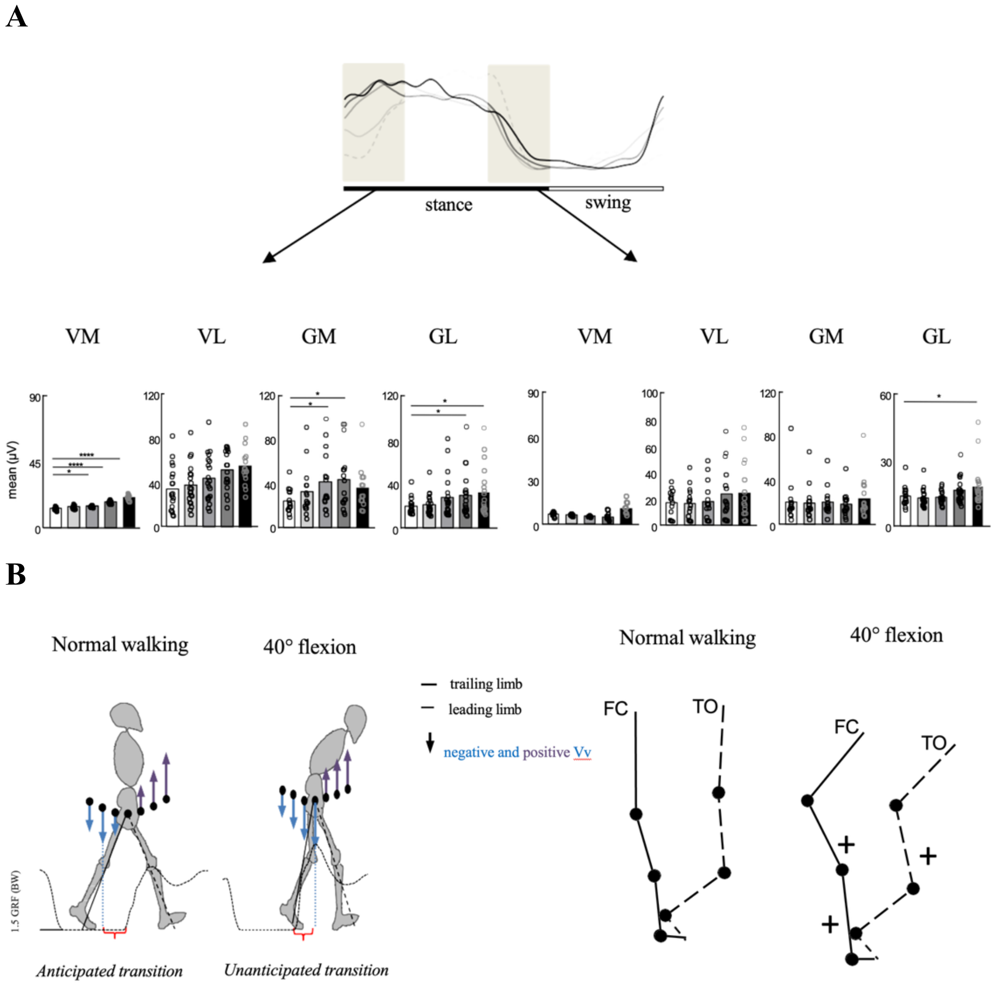

Fig. 2.7. *Initial and final muscular activity and schematic summary of the study results. **(A)** Top: typical traces of the amplitude of the activation of the MG at each condition. The grey areas indicate the initial and final 20% of the stance phase during a stride. Bottom: mean activation of the VM, VL, GM and GL during the initial (left) and the final (right) 20% of*

*the stance phase. Other indications as in Fig. 2.2. (**B**) Left: schematic representation of the anticipated and unanticipated transition, defined by the time of the redirection velocity of the CoM relative to FC, during normal walking and with 40 deg flexion. The arrows represent the vertical velocity vector of the CoM. At the bottom, the corresponding vertical forces of the front legs during the double contact phase are represented, as well as the delay between FC of the front leg and the minimum vertical velocity (red brackets). Right: schematic representation of the muscle activation at the FC and TO with normal walking and 40 deg trunk flexion. The plus signs (+) indicate a greater muscular activation than during normal walking.*

Discussion

This study aimed to determine the effect of trunk flexion on the neuromuscular control of gait. Specifically, we analyzed the effect of trunk flexion on lower-limb kinematics, the step-to-step transition of walking and, with it, its impact on lower-limb muscle activation patterns during walking. The results are discussed in the context of the biomechanical principles underlying foot-support interactions during step-to-step transitions in bipedal upright walking.

Given that the trunk and head segment constitute approximately 65% of the body mass (Dempster, 1955), walking with the trunk anteriorly flexed modifies the position of the CoM, which is displaced forward and slightly downward (Müller et al., 2017). As a result, even a small change in trunk orientation can influence the mechanical demands of walking (Figs 3.3 and 3.4; (Aminiaghdam et al., 2017; Hora et al., 2024; Müller et al., 2017), in turn affecting the neuromechanical control of locomotion. As illustrated in Fig. 2.7B, trunk flexion alters the spatial relationship between the CoM and the FC point of the back leg at the end of the single stance (Warrener et al., 2021) resulting in different leg orientation (Aminiaghdam et al., 2016). This results in a reduction of the vertical and fore–aft forces under the back leg during propulsion (Figs 3.4C and 3.7B; (Aminiaghdam et al., 2016; Grasso et al., 2000). Additionally, the GRF is more vertical at FC because Ffront increases at weight acceptance without changes in $F_{f,brake}$ (Fig. 2.4C). Similar modifications of vertical impulses have been observed in older adults (Delabastita et al., 2021; Dewolf et al., 2022; Meurisse et al., 2019b) and in trunk flexion-induced gait analysis (Müller et al., 2017). We documented here that it has a clear impact on step-to-step transition: an unanticipated transition was revealed at 20, 30 and 40 deg of trunk flexion (Figs 3.1B, 3.4B and 3.7B), accompanied by a reduction in the mechanical recovery of walking at 40 deg of trunk flexion (Fig. 2.5B). The unanticipated transition occurs when $V_{v,min}$ appears after FC. This absence of modification of transition at 10 deg (and in turn of mechanical work; Fig. 2.5B) may be due to the minimal changes in the energy fluctuations (Fig. 2.5A). Instead, with a greater trunk inclination, the vertical displacement of the CoM starts to increase and the modification of the ratio between potential and kinetic work over one stride (W_p/W_k) induces a reduction of recovery between energies (Cavagna et al., 1976, 1977). The detailed results of W_p/W_k are shown in Fig. S2.1.

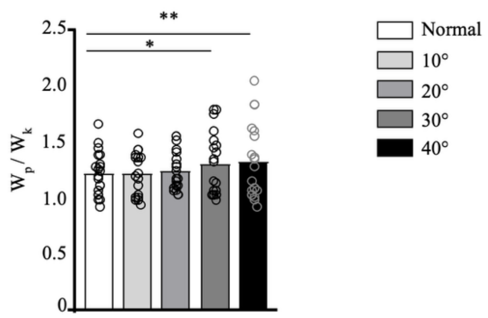

Fig. S2.1. *Average comparison of walking conditions with trunk flexion: the ratio of potential and kinetic work over one stride (W_p/W_k) with increasing trunk flexion. The bars represent the average, and the circles represent the mean values for each condition participant. The asterisks (*) denote post hoc comparisons between the normal condition and the other conditions (*$p < 0.05$, ** $p < 0.01$, *** $p < 0.001$, **** $p < 0.0001$).*

As for the modification of muscle activation during the stride (Figs 2.1 and 2.6), it is noteworthy that with increasing trunk flexion, the activation of extensor muscles generally increases, except for GL and GM (Fig. 2.6B). A similar result has also been described recently in crouched walking (Hora et al., 2024), showing lower increases of activation in the lower leg muscles with trunk inclination compared with the thigh muscles. As suggested by Hora et al. (2024), the lower increase in activation of ankle muscles might reflect the smaller changes in ankle joint kinematics compared with the hip and knee joints (Fig. 2.3A,B). When specifically examining the activation of the extensor muscles in the front leg, VM increases its activity mainly during the initial 20% of the stance phase (Fig. 2.7A), confirming the greater contribution of VM during weight acceptance. However, during the initial 20% of stance, GM and GL also increase with increased trunk flexion, suggesting an antigravitational role (Hamner et al., 2010; Honeine et al., 2013) when the collision with the ground of the front leg becomes more important. This increase in distal extensor activation is related to their earlier activation (CoA; Fig. 2.6B, bottom) during the walking cycle. In addition, we noticed a change in the period during which muscle activity remains consistently above half of its maximum level (Fig. 2.6B). Here, we evidenced a higher FWHM with an increasing trunk inclination (Fig. 2.6B, middle) in proximal (VM and VL) and distal muscles (GM and GL). These prolonged activations probably suggest the need for enhanced stability and control as the body adapts to the forward trunk inclination. Similar compensatory mechanisms are observed in the gait pattern of patients with balance deficit, such as cerebellar ataxia (Martino et al., 2014). In summary, when the lower limbs are more flexed and the CoM is shifted forward and downward, increased muscle activity is required to generate the necessary joint torque to accomplish the walking task.

This is also related to the more flexed lower-limb joint during stance (Fig. 2.3A,B; (Aminiaghdam et al., 2016)). Indeed, as trunk flexion increases, the participants exhibit a linear increase in hip, knee and ankle (dorsi-) flexion, while the ROM of these joints remains consistent throughout the step (Fig. 2.3A,B). This most likely induces muscle compensation reminiscent of crouched walking (Grasso et al., 2000). Indeed, Grasso et al. (2000) observed that imposing greater knee flexion during stance leads to a higher muscle activation of lower-limb muscle. As for walking with a flexed trunk, the increment of muscle activity can be related, among others, to the lower mechanical energy recovery through the pendular energy exchange (see figure 7 in Li Y. et al., 1996). Additionally, Grasso et al. (2000) also observed that the time sequence of activation of different muscles varied substantially as a function of imposed posture. In particular, as in the present study, a greater activation of the knee extensor (VL) was observed at the end of the stance phase, and a greater activation of the ankle extensor (GM) at the beginning of the stance phase, consistent with our findings. In contrast, our study found greater activation of the knee extensor (VM) at the beginning of the stance phase (Fig. 3.7A).

This suggests that trunk inclination and the lower-limb kinematics change in parallel and cannot be studied separately. Similar observations were previously made by Grasso et al., (1999) who showed that in Parkinson's disease patients in the apomorphine OFF condition, the flexion of the trunk is paralleled by the flexion of the legs. In the ON condition, the trunk and the limbs extend together. These authors proposed that the basal-ganglia circuitry may provide a spatiotemporal framework for the control of both posture and movement in a gravity-based frame.

Also, a recent study (Mesquita et al., 2023) suggests that altering the foot interaction with the ground affects the walking pattern similarly to trunk flexion. By imposing different foot positions during stance (e.g. maintaining a flat-foot during the whole stance phase), the authors described that deviating from the classical heel- to-toe pattern of walking leads to a higher F_{front}, a smaller F_{back}, an anticipated activation of distal extensor muscles and a higher activation of proximal muscles. They argued that the specific use of the heel-to-toe rolling pattern allows delay of the activation of distal extensors relative to proximal extensors, ensuring a smooth step-to-step transition. This is a specific adaptation of the locomotor apparatus to bipedalism. Indeed, in quadrupeds, the step-to-step transition is not as crucial as in bipeds because the mechanical cost of the redirection is reduced in proportion to the number of contacts used to achieve a given redirection of the CoM (Ruina et al., 2005). In addition, the change of the hindlimb configuration at the end of stance (knee flexed) may allow them to propel themselves forward using primarily the proximal extensors. Interestingly, deviating from the erect posture of the trunk, another specific adaptation of the locomotor apparatus to bipedalism, results in similar modifications of walking pattern.

These indications provide arguments to the fact that the erect posture has evolved to optimize gait, allowing the CoM to vault over the supporting straight limb. Indeed, based on the fluctuation of the mechanical energy of the CoM, normal walking is often compared to an inverted pendulum model based on the fluctuation of the mechanical energy of the CoM (Cavagna &

Kaneko, 1977; Willems et al., 1995). Maintaining a flexed trunk during walking potentially affects this pendulum-like mechanism as it modifies the position of the CoM and the limb posture during stance. Here, when comparing normal walking and walking with a 40 deg trunk inclination, we found a significant decrease of the energy transduction between E_p and E_k (%R) and with it a medium effect (ES=0.56) on the mechanical work done to move the CoM (W_{ext}) (Fig. 2.5B). Furthermore, the pattern of walking with imposed trunk flexion is more similar to that of bipedal chimpanzee gait than to normal walking. Indeed, the kinematics and mechanics of walking with imposed trunk flexion overlap widely with the gait pattern of chimpanzee bipedal walking. Chimpanzees walk with a crouched posture with flexed limbs (Johnson et al., 2022) and similar CoM mechanics and recovery rates can be observed for chimpanzee bipedal walking (Demes et al., 2015). Understanding the link between posture and gait mechanics across different species may help researchers better interpret the effects of evolution on bipedal walking.

More surprisingly, the opposite was observed when comparing normal with walking with a 10 and 20 deg flexed trunk, i.e. an increase in %R and a decrease in W_{ext}. This last result suggests that the recovery of energy only gives partial information concerning the metabolic cost of walking as the collisional loss occurring at contact is more likely to be a better determinant of the cost of locomotion. Nevertheless, it allows a better understanding of the mechanical– bioenergetic interaction of walking. In general, humans tend to prefer economical walking patterns. However, (L. C. Hunter et al., 2010) showed that when walking on a shallow downhill slope, their participants did not take optimal advantage of the propulsion provided by gravity to decrease energetic costs, but instead preferred a more stable and costlier gait pattern. Here, it could be that, because of the shift in the hip relative to the foot, a 10 and 20 deg trunk flexion allowed the participant to take advantage of the propulsion provided by gravity and thus to reduce the mechanical cost of walking at the expense of stability. When the trunk becomes more flexed, it can be observed that E_k and E_p become more in phase after foot contact (Fig. 2.5A). This suggests that the pendulum-like exchange is progressively reduced, and other mechanisms such as elastic energy storage potentially come into play (Dewolf et al., 2016).

In conclusion, this research sheds light on the impact of trunk flexion on gait dynamics and its neuromuscular control. The study reveals that even subtle changes in trunk orientation, as observed in older adults (Dewolf, Sylos-Labini, Cappellini, Ivanenko, et al., 2021; Dewolf et al., 2022), lead to adjustments in lower limb kinematics, step-to-step transitions, GRFs and lower- limb muscle activation patterns (Fig. 2.6B). Notably, increased trunk flexion introduces mechanical challenges as a result of the modification of CoM position (Müller et al., 2017). These findings underline the complex interplay between mechanics and neural control of gait, enhancing our understanding of human bipedal walking and its biomechanical underpinnings.

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Chapter III

The Role of Physical Activity in Mitigating Age-Related Changes in the Neuromuscular Control of Gait

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List of Abbreviations

BW:	Body weight
CI:	Co-activation index
CoA:	Center of activity
CoM:	Center of mass
CoP:	Center of pressure
E_k:	Kinetic energy of the CoM
EMG:	Electromyography
E_p:	Potential energy of the CoM
ES:	Erector spinae
FWHM:	Full width at half maximum
GM:	Gastrocnemius medialis
GL:	Gastrocnemius lateralis
Gmed:	Gluteus medius
GRF:	Ground reaction force
IMVT:	Isometric maximal voluntary torque
MN:	Motoneuron
OA:	Older adults
RF:	Rectus femoris
RMS:	Root mean square
SOL:	Soleus
ST:	Semitendinosus
TA:	Tibialis anterior
TFL:	Tensor fasciae latae
tsDCS:	Trans-spinal direct current stimulation
VL:	Vastus lateralis

VM:	Vastus medialis
$V_{v, min}$:	Minimum vertical velocity of the CoM
W_{ext}:	External mechanical work
YA:	Young adults

Abstract

Background: Exercise induces neural and muscular adaptations, improving muscle mass and function in older adults. We investigated its impact on gait neuromuscular control in young and older adults, classified as more active (young: n=15, 5185 ± 1471 MET-min/week; old: n=14, 6481 ± 4846 MET-min/week) or less active (young: n=14, 1265 ± 965 MET-min/week; old: n=14, 1473 ± 859 MET-min/week). Isometric maximal voluntary torques were assessed for proximal (knee) and distal (ankle) extensors, and muscle mechanical properties of these muscles were assessed using Myoton. Gait was analysed using ground reaction forces, motion capture, and electromyography. Less active older adults exhibited shorter steps, higher mechanical cost, and greater collision at heel strike. These differences were linked to altered neuromuscular control, wider activation of lumbar and sacral motor pools, different activation timing, and reduced muscle-tendon stiffness. Our findings highlight that physical activity preserves neuromuscular control, muscle mechanical properties, and gait efficiency, mitigating age-related decline.

Introduction

With extended life expectancy, the loss of mobility of older adults has dramatic individual and societal impacts. Among the many factors contributing to frailty (e.g., metabolic, hormonal, and immunological changes), the loss of muscle mass and strength (Akima et al., 2001; Hyatt et al., 1990), alterations in connective tissue properties (Graça et al., 2023), and neurodegenerative changes (Rygiel et al., 2016b) play a key role in the decline of mobility and functional capacity in older adults. These neuromuscular alterations result in an increased metabolic rate of locomotion in older adults, associated with a loss of independence and an increased risk of morbidity and mortality (Cooper et al., 2010, 2011; Guralnik et al., 1995; Studenski et al., 2011).

Several age-related modifications of the muscle and tendon tissues have been described, such as loss of muscle fibres (Lexell, Henriksson-Larsén, et al., 1983; Lexell, 1995; McPhee et al., 2018), a shift in muscle fibre type distribution (Williamson et al., 2000), modification of the tendon and muscles cross-sectional area (Lexell & Taylor, 1991a; Stenroth et al., 2012) or changes in the stiffness and viscoelastic properties of tendon and muscles (Akagi et al., 2015; Çekok et al., 2024; Lim et al., 2019; Şendur et al., 2020; Stenroth et al., 2012), associated with histological changes such as impaired connective tissues or myosteatosis. The decrease in muscle-tendon unit performance has an impact on mobility of the older adults. For example, older adults redistributed lower extremity joint moment and power compared with young people in walking (DeVita & Hortobagyi, 2000; Silder et al., 2008), so-called biomechanical plasticity. The scientific community (Boyer et al., 2023; Delabastita et al., 2021; DeVita & Hortobagyi, 2000; Franz, 2016; Franz & Kram, 2013; Gueugnon et al., 2019a; Kerrigan et al., 2000; McGibbon, 2003; Winter, Patla, et al., 1990) most often points to a reduction in mechanical power generated by the plantar flexor muscles during the push-off phase of walking as the hallmark biomechanical ageing features of gait.

The decline in mobility and functional capacity in older adults is influenced by multiple factors, including muscle weakness, changes in connective tissue properties, and neurodegenerative processes. While the reduction in ankle muscle contractile performance significantly affects mobility (Marcucci & Reggiani, 2020), other neuromuscular factors also play a crucial role (Aagaard et al., 2010; S. K. Hunter et al., 2016). In particular, age-related changes in spinal motoneuron activity impact force production and movement control. Older adults exhibit altered motoneuron recruitment and firing characteristics (Hassan et al., 2021a; Orssatto et al., 2022), leading to reduced muscle function. Additionally, age-related modifications in spinal segmental output—such as broader motoneuron activation patterns, increased co-activation, and differential impairment of caudal versus rostral segments—may contribute to changes in locomotor coordination and efficiency (Dewolf et al., 2021a, 2021b; Nùñez-Lisboa et al., 2023). Decoding these neuromuscular changes is essential for understanding their contribution to gait adaptations in aging and may be essential for linking the modification of output with neuromuscular degeneration and the firing characteristics of motoneurons.

High physical activity level is crucial for enhancing physical fitness by increasing muscle strength, and cardiopulmonary function and stimulating metabolic activity (Chen et al., 2021; da Cunha et al., 2015; el Hadouchi et al., 2022; Kraemer, Adams, et al., 2002; Markov et al., 2022; Pour et al., 2017), decreasing risk of disability (Reid & Fielding, 2012). Regular physical activity plays a role in increasing/maintaining muscle mass, strength, power, and functional capacity in older adults, but it also modify age-related changes in motor unit structure and function across the life span (Cogliati et al., 2020; S. K. Hunter et al., 2016). For example, exercise promotes various metabolic responses and morphological reconfigurations in the neuromuscular junction of older adults(Rogers & Nishimune, 2017), decelerating or even reversing neuromuscular junction degeneration (Q. Wang et al., 2024). On the contrary, less regular physical activity among older adults may exacerbate age-related differences between young and older adults, for example in muscle activation (Harridge et al., 1999). However, enhancing physical capacity alone may not be sufficient to mitigate the age-related decline of the neuro-muscular system. Indeed, increasing the physical activity of older adults via resistance training fails to directly translate to improved propulsive power generation in walking (Beijersbergen et al., 2013). Similarly, Brach et al. (2013) assert that multicomponent impairment-based walking exercises, while beneficial for strength, flexibility, and endurance, don't always lead to better walking ability. Also, a higher level of physical activity in older adults did not mitigate the age-related modification of kinematic coordination and distal-to-proximal redistribution (Boyer et al., 2012), referring to the shift in mechanical work and power generation from distal joints (such as the ankle) to more proximal joints (such as the hip) during walking (Franz, 2016), suggesting that there remains a gap in understanding the biomechanical adaptations that physical activity levels may induce in gait mechanics across aging (Beijersbergen et al., 2013).

While a large body of literature has focused on the role of physical activity in preserving mobility, it is still difficult to disentangle the influence of biological aging from that of physical activity on the subsequent changes in the neuromuscular system. To address this question, we recorded data from both older and younger participants, divided according to their weekly physical activity levels. After evaluating the static balance, isometric strength and the mechanical properties of ankle and knee extensors, we then compared the mechanics and neuromuscular control of gait. The purpose of this study was to shed light on contributors to the age-related changes in biomechanical, physiological, and motor control adaptations of gait. We first hypothesized that a clear effect of aging on muscular mechanical properties, maximal isometric forces and neuromuscular control of gait would be observed. Also, we tested the hypothesis that more active older individuals exhibit a “slowing down” in the natural aging process of the neuromuscular system.

Materials and methods

Participants

Fifty-nine healthy young and older adults (♂ = 39, ♀ = 20) participated in this study. Both age groups were divided into two groups (less and more active), according to the median level of physical activity, measured through the Global Physical Activity Questionnaire (GPAQ) (Fig. 3.1). A modified z-score using the median and Median Absolute Deviation (MAD) was applied to provide a robust classification of participants into "Less Active" and "More

Active". The four groups' characteristics are presented in Table 1. Sample size was based on an a priori power analysis (effect size = 0.4, power >80%) from Dewolf et al.(A. Dewolf, Sylos-Labini, Cappellini, Ivanenko, et al., 2021), based on unpublished correlation between physical functioning and step period. Our test indicated that 12 subjects per group would be sufficient. The inclusion criteria were the following: ability to walk a kilometre, no locomotor system injury complaints and no previous history of neurological disorders. All participants provided informed consent, and the procedures used for this study were approved by the ethics committee of the Université Catholique de Louvain (Belgian Registration Number: B403201524765) and adhered to the Declaration of Helsinki.

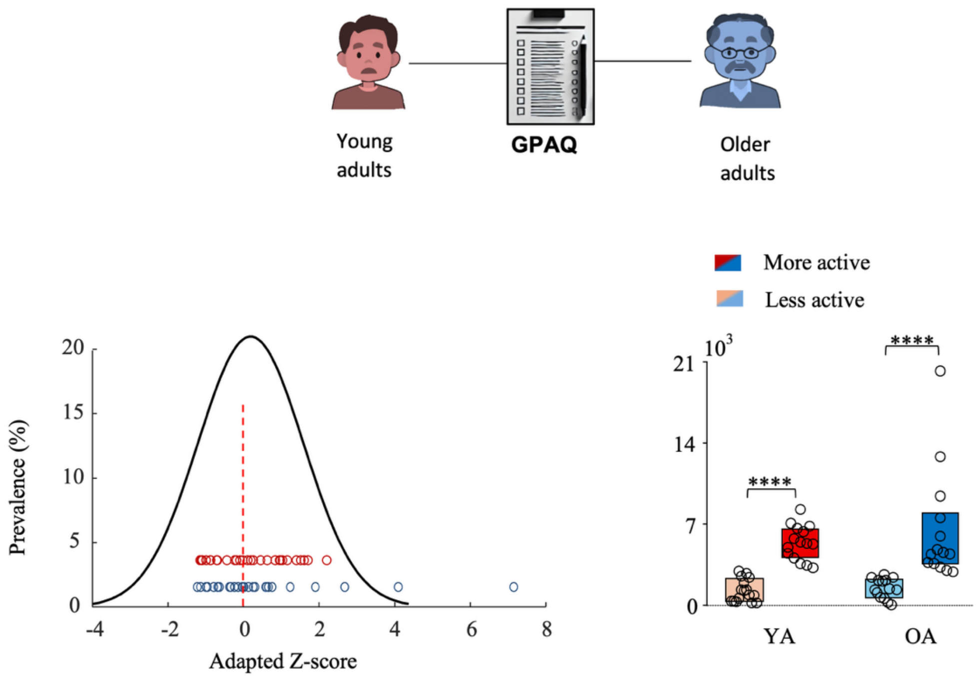


Fig. 3.1. *Group classification based on physical activity levels. Left panel: Distribution of physical activity levels in MET-min/week in young (red dots) and older adults (blue dots), classified using the Global Physical Activity Questionnaire (GPAQ). Older adults had a broader distribution of activity levels compared to the younger group. A modified Z-score was applied to normalize activity levels. In this way, participants with a positive/negative adapted Z-score were classified as more/less active. Right panel: Boxplot displaying the average MET-min/week scores per group. The statistical significance of the differences*

*between groups was assessed using the Kruskal-Wallis test. The asterisks indicate post hoc comparisons between groups (*p < 0.05, **p < 0.01, ***p < 0.001, ****p < 0.0001).*

Table 3.1 Subject characteristics grouped based on their age and level of physical activity.

	Young adults		Older adults		
variable	*Less active*	*More active*	*Less active*	*More active*	
n	16	15	14	14	
Age (years)	26.7 ± 4.0	25.5 ± 2.0	73.1 ± 5.5	69.5 ± 2.5	+
Height (cm)	178.8 ± 8.6	179.4 ± 8.8	169.7 ± 9.7	174.0 ± 9.3	+
Weight (kg)	73.2 ± 10.8	72.9 ± 8.5	72.2 ± 15.9	76.3 ± 13.6	
BMI (kg m^{-2})	23.1 ± 4.4	22.7 ± 1.9	24.9 ± 3.6	25.0 ± 3.0	
Knee extensor strength (N m)	194.0 ± 48.8	203.7 ± 54.4	119.9 ± 40.5	150.7 ± 37.5	+
Ankle extensor strength (N m)	47.8 ± 25.2	65.0 ± 19.7	30.7 ± 12.6	35.8 ± 10.3	+ £
Met-min/week	1265 ± 965	5185 ± 1471	1473 ± 859	6481 ± 4846	£

***Table N° 3.1** presents the characteristics of the young and older adult participants, grouped as either less active or more active based on their MET-min/week values. The variables include age (years), height (mt), weight (kg), body mass index (BMI) (kg/mt^2), knee extensor strength, and plantar flexor strength (N.m). The values are reported as means with corresponding standard deviations (±). Significant effects of age, physical activity level (Pal), and their interaction (Age*Pal) are noted with the symbols +, £, and $, respectively.*

Experimental procedure and data collection

Postural control and gait analyses were conducted using an instrumented treadmill (H/P Cosmos-Stellar treadmill, Germany, belt surface: 1.6 x 0.65 m) equipped with four force transducers (Arsalis®, Belgium). The transducers were placed under the body of the treadmill, the force transducers measured the three components of ground reaction force (GRF) exerted by the treadmill belt under the foot (Willems & Gosseye, 2013): F_v, F_f, and F_l, respectively the

vertical, fore-aft, and lateral components of the GRF. Data were sampled at a frequency of 1000 Hz.
Electromyographic (EMG) muscle activity from 12 muscles on the right side of the body was captured at a frequency of 2048 Hz using a Delsys Trigno Wireless System (Boston, MA). The muscles registered were the ES at the L2 level, the TFL, the Gmed, the vastus medialis (VM), the vastus lateralis (VL), the RF, the TA, the ST, the BF, the GL, the GM, and the SOL. Prior to electrode placement, the skin was prepared by shaving and cleaning with fine sandpaper, ether, and alcohol following guidelines from SENIAM (the European project on surface EMG—seniam.org). To ensure accurate placement, the muscle bellies were located by palpation, and electrodes were aligned with the main direction of the muscle fibres (66). Electrode placement and signal quality were verified through visual inspection of EMG signals while participants contracted each muscle. EMG data and kinetic measurements were synchronized using a Delsys Trigger Module. Bilateral, full-body three-dimensional kinematics were recorded using a Qualisys system (Gothenburg, Sweden) equipped with 14 Oqus 600+ cameras and 4 Miqus M3 cameras placed around the treadmill. Participants were equipped with 20 retro-reflective markers, placed bilaterally on specific anatomical landmarks: neck, shoulders, wrists, posterior superior iliac spines, anterior superior iliac spines, greater trochanters, knees, malleoli, elbows, and the fifth metatarsal bones, according to the Qualisys sports marker set disposition.

The participants were first asked to stand on an Instrumented treadmill to perform a static balance test. They stand with their feet together, eyes closed, for 30 seconds each. During the test, they were instructed to maintain their arms relaxed at their sides, avoid compensatory movements, and stand as still as possible. The forces exerted by the ground on the feet of the participant during standing were measured on the instrumented treadmill. From the GRF, the lateral and fore–aft positions of the centre of pressure (CoP) were computed as follows:

$$CoP_x = \frac{-M_y - hF_x}{F_z}$$

$$CoP_y = \frac{-M_x - hF_y}{F_z}$$

Where F_x, F_y, and F_z are the lateral, fore-aft, and vertical components of the GRFs; M_x and M_y are the moment components in the force transducer coordinate system; and h is the vertical distance between the force transducers and the tread surface (A. H. Dewolf et al., 2018). The range of CoP_x and CoP_y oscillation was then determined.

Following the balance test, biomechanical properties of the *gastrocnemius medialis* (MG) and *rectus femoris* (RF) were measured using a handheld, non-invasive MyotonPRO device (Myoton AS, Tallinn, Estonia). Specifically, we measured these muscles 'dynamic stiffness and dampening (logarithmic decrement) by applying a mechanical impulse and recording the muscle's response (Garcia-Bernal et al., 2021b). We measured the biomechanical properties of the *rectus femoris* (RF) and *gastrocnemius medialis* (GM) at locations where reliability has been previously confirmed. Specifically, the RF was measured at two-thirds of the distance between the anterior superior iliac spine (ASIS) and the superior pole of the patella(Agyapong-Badu et al., 2016) and the GM was measured at the most reliable site identified by Nguyen et al(Nguyen et al., 2024). Five successive measurements were taken for each muscle, while participants were standing still. Subsequently, the participants were asked to walk wearing their shoes on an instrumented treadmill at 1.11 m s^{-1} (4 km h^{-1}). The selected walking speed was similar to the one used in our previous studies on older adults (Dewolf, Sylos-Labini, Cappellini, Ivanenko, et al., 2021; Dewolf, Sylos-Labini, Cappellini, Zhvansky, et al., 2021a; Dewolf et al., 2022; Meurisse et al., 2019b) and was reported as the comfortable and economical average walking speed based on the net energy consumption (Cavagna & Kaneko, 1977). Before starting data collection of walking, participants familiarized themselves with treadmill walking through multiple trials with verbal feedback from the experimenter. After the familiarization period, we gradually ramped up the treadmill to 4 km h^{-1} and then waited at least 30 seconds before starting the kinematics, GRF and electromyographic data acquisition. On average, 28.1 ± 2.9 (mean ± SD) strides were analysed per participant.

After completing the walking trial, the participants performed Isometric Maximal Voluntary Torque (IMVT) measurements for the knee and ankle extensors. The isometric strength tests were performed using a 1D strain gauge (Arsalis®, Belgium) to measure torque(Mesquita et al., 2020), on a custom-designed table equipped with adjustable straps. The IMVT protocol began with a warm-up, during which participants performed incremental repetitions. Following the warm-up, the participants were instructed to exert their maximum effort for 6 seconds. Three trials were recorded, and the highest torque was analysed. The IMVT data were sampled at 500 Hz. For the knee extensor torque measurements, participants were seated with their hips, knees and ankles positioned at a 90° angle, secured with adjustable straps, and a rigid rope was fixed horizontally between the ankle and the gauge fixed on the wall (Nordin et al., 2020). For ankle extensor torque, the participants were seated in the same position, but with the shank fixed vertically using adjustable straps. A rigid rope was fixed vertically between the fifth metatarsal and the strain gauge, fixed on the wall. Participants were instructed to avoid trunk flexion to ensure accurate torque measurements. The protocol and setup were adapted from Dragoi et al. (Dragoi et al., 2022).

Data analysis

Division of the stride: The first instant of FC and the toe-off (TO) events were estimated from the displacement of the centre of pressure on the belt (Meurisse et al., 2016b). The FC and TO were automatically identified based on the computation of d_{path}, which represents the difference between the CoP trajectory and a straight-line approximation connecting its initial and final positions within a step cycle (Meurisse et al., 2016b). A stride was delimited by two successive right FCs. A step was defined as the interval between one leg's FC and the contralateral leg's FC. Stance phases were measured as the time between FC and TO of the same leg.

Posture: The lateral (CoP_x) and anteroposterior (CoP_y) displacement of the centre of pressure (CoP) were analysed for 30 seconds standing with closed eyes. A 4th-order low-pass Butterworth filter was applied with a 20 Hz cut-off frequency (A. H. Dewolf, Ivanenko, Mesquita, & Willems, 2021). Specifically, the ranges of CoP displacement in both directions (CoP_x and CoP_y) were determined as two standard deviations (±1 s.d.) of the time series, following the approach described by Dewolf et al., 2021a, providing a measure of the postural oscillation around the average position of the CoP.

Walking kinetics: Data were recorded at a sampling rate of 1000 Hz. The fore-aft and vertical velocity of the CoM were determined from the fore-aft and vertical components of the GRF using the procedure described in detail in Dewolf et al. (2016). In short, the fore-aft acceleration of the CoM was calculated as $a_f = F_f/m$, where m is the subject's body mass. The vertical acceleration of the CoM was calculated as $a_v = (F_v - m\,g)/m$, where g is the acceleration of gravity. The vertical (V_v) and the forward velocity (V_f) of the CoM were calculated by time-integration of a_v and a_f, respectively, plus an integration constant, which was computed so that the average velocity over a stride was equal to zero. The vertical and forward displacements of the CoM (S_v and S_f, respectively) were then computed by time-integration of V_v and V_f.

The energy of the CoM (E_{com}) was computed as the algebraic sum at each instant of its gravitational potential energy ($E_p = m\,g\,S_v$) and its kinetic energy ($E_k = ½\ m\ (V_f^2 + V_v^2)$). The work done to move the CoM relative to the surroundings (W_{ext}) was then computed as the sum of the positive increments of E_{com} (Cavagna et al., 1976). The energy transduction between E_p and E_k was estimated from the relative amount of energy saved over a step (% *recovery*):

$$\%recovery = \frac{\mathrm{W_k} + \mathrm{W_p} - \mathrm{W_{ext}}}{\mathrm{W_k} + \mathrm{W_p}} * 100$$

where W_k and W_p correspond to the sum of the positive increments of E_k and E_p, respectively. The external power P_{com} was computed as $P_{com} = dE_{com}/dt$. During a walking stride, the P_{com} time-curve typically shows one major negative peak (W^-) and two major positive peaks (W^+_1 and W^+_2), known as collision and propulsion, respectively (A. H. Dewolf, Ivanenko, et al., 2019b). The maximum values of these peaks were measured for all trials.

Kinematics: Kinematic data were recorded at a sampling rate of 240 Hz and later oversampled during post-processing to 1 kHz, to match the kinetic sampling frequency. The elevation angle of each segment (trunk, thigh, shank, and foot), defined as their orientation relative to the vertical (Borghese et al., 1996; A. H. Dewolf et al., 2018) were computed, with 0° corresponding to alignment with the vertical axis during walking. Joint angles for the hip, knee, and ankle were calculated based on the relative orientation between adjacent segments. Each stride was interpolated over 400 points, time-normalizing the data across all trials for each subject.

EMG: The 12 raw EMG signals were resampled at 1000 Hz to match the sampling frequency of the kinetic data. The signals were high-pass filtered at 30 Hz, then rectified, and subsequently, low-pass filtered using a zero-lag 4th-order Butterworth filter at 10 Hz. Additionally, a semi-automatic artefact detection procedure was implemented. A notch filter at 50 Hz was applied to remove frequency-related artefacts, such as power line interference and its harmonics. Then, gait cycles containing evident artefacts, such as signal loss or saturation due to sensor detachment, were manually excluded following visual inspection. Finally, an outlier detection method was applied: individual gait cycles were flagged for further review if their correlation coefficient with the ensemble-averaged EMG envelope fell below 0.6, indicating substantial deviation from the typical signal pattern (Zhvansky et al., 2022). The time scale was normalized by interpolating each gait cycle to 400 points. The root mean square (RMS) of each EMG signal was then computed as a measure of overall muscle activation amplitude across the gait cycle. For each condition, the FWHM was determined as the duration during which the EMG activity exceeded half of its maximum value (Martino et al., 2014; Santuz et al., 2020). The centre of activity (CoA) of each EMG signal was measured as an estimation of the timing of activation. The CoA during the gait cycle was calculated using circular statistics (Batschelet, 1981) and plotted in polar coordinates (polar direction denoted the phase of the gait cycle, with angle α that varies from 0 to 360°).The CoA of the EMG waveform was calculated as the angle of the resultant vector (i.e., the first trigonometric moment) pointing to the center of mass of the circular distribution of EMG activity relative to the gait cycle (Martino et al., 2014), computed as:

$$A = \sum_{i=1}^{400}(\cos \alpha_i X\ EMG_i),$$

$$B = \sum_{i=1}^{400}(\sin \alpha_i X\ EMG_i),$$

$$CoA = tan^{-1}(B/A)$$

where EMG_i represents the EMG amplitude at each phase of the gait cycle. The CoA was preferred, as identifying a clear peak of activity was impractical for most muscles (Martino et al., 2014).

The EMG activity was then mapped onto the rostro-caudal positions of the MN pools in the human spinal cord, ranging from segments L2 to S2. This mapping followed the myotomal charts of Kendall et al. (F. Kendall et al., 2005), based on the approximate rostro-caudal location of MN pools innervating different muscles in the human spinal cord. In general, each muscle is innervated by several spinal segments. To reconstruct the output pattern of any given spinal segment S_j, all rectified EMG waveforms corresponding to that segment were averaged as:

$$S_j = \frac{\sum_{i=1}^{n_j} k_{ij} \cdot EMG_i}{n_j},$$

where n_j is the number of EMG_i waveforms corresponding to the j^{th} segment, k_{ij} is the weighting coefficient for i^{th} muscle (from Kendall's chart). The assumption implicit in both methods is that the rectified EMG provides an indirect measure of the net firing of MNs of that muscle in the spinal cord. To compute the total motor output for each condition, the motor output patterns across the gait cycle were summed across the lumbar, sacral and all spinal segments. The mean activation of the lumbar (L2 to L5) and sacral (S1 to S2) segments was computed by averaging the motor output patterns for each region. Finally, the FWHM and the CoA were calculated for both lumbar and sacral segments. The co-activation index (CI) was assessed between the lumbar and sacral segments using the following formula (Mari et al., 2014; Rudolph et al., 2000):

$$\mathrm{CI} = \frac{\sum_{j=1}^{400}\{[\mathrm{lumbar_j}+\mathrm{sacral_j}]/2\}\mathrm{x}[\mathrm{Lumbar_j}/\mathrm{sacral_j}]}{400},$$

where sacral and lumbar represent the mean activation of the segments during the stride. The CI was then averaged over the entire gait cycle (j = 1:400), providing an estimate of the co-activation over the entire cycle. High CI represent a high level of activation of both spinal segments across a large time interval, whereas low CI indicate either a low-level activation of both segments or a high-level activation of one segment along with low-level activation of the other one. Spinal-level co-activation provides a functional representation of neuromuscular output, as previously used to assess age-related co-activation(Dewolf et al., 2021b).

IMVT: A custom-made MATLAB script was utilized to accurately determine the peak force generated during the 6-second maximal voluntary contraction (MVC) trial. The script processed the raw data, applying a low-pass filter at 20 Hz. After filtering, the peak torque was identified as the maximum force maintained for at least half a second. This approach avoids the detection of a short peak of force, providing a precise measurement of the participant's maximal torque output during the isometric test. The lever arms were measured using anthropometers.

Statistics

For each dependent variable (stance duration, step length, external work of the CoM, recovery, peak negative power, peak positive power 1 and 2 of the CoM, amplitude of CoP_x and CoP_y, elevation angles, joint kinematics, foot angle at touchdown, peak knee flexion, muscle activation metrics (RMS, FWHM, CoA), spinal motor outputs, co-activation index, myoton properties, and IMVT torque measurements), the normality of the residuals was visually assessed using QQ plots. For variables that did not meet the normality assumption, a log10 transformation was applied, and normality was confirmed afterwards. A mixed-effects model was performed with age (young *vs.* older adults) and physical activity level (less *vs.* more active) as fixed effects. A post-hoc Fisher LSD multiple comparisons test was used to evaluate differences between age groups and levels of physical levels. The interaction between the fixed effect was reported if it was significant. All statistical analyses were conducted using SPSS 23 (IBM, USA) with an alpha level of 0.05.

Results

Some small differences in subject characteristics were found between groups, presented in Table 3.1. Subjects physically more active were slightly younger than less active ones ($F_{1,55}$ = 2144.0, p< 0.001; $F_{1,55}$ = 6.0, p= 0.017). Older adults were on average shorter than young adults ($F_{1,55}$ = 8.6, p= 0.004). Regarding the subject's weight, no significant differences were observed between groups (p> 0.743). However, the BMI was slightly lower in more active individuals ($F_{1,55}$ = 4.0, p= 0.049), while age ($F_{1,55}$ = 0.3, p= 0.579) had no effect. More importantly, no effects of age ($F_{1,55}$ = 0.9, p= 0.334) or interaction ($F_{1,55}$ = 0.4, p= 0.511) were observed on the classification using the estimated MET-min/week.

Muscular properties

The maximal isometric torques were affected by age (Fig. 3.2A). Indeed, both the knee and ankle extensor strength were smaller in older adults compared to young participants (knee: $F_{1,55}$ = 27.5, p< 0.001; ankle: $F_{1,55}$ = 26.9, p< 0.001). No difference was observed in knee extensor torque as a function of physical activity level ($F_{1,55}$ = 2.7, p = 0.107). However, an effect of level of activity was observed on plantar flexor torque ($F_{1,55}$ = 6.2, p= 0.020), with the more active participants being stronger.

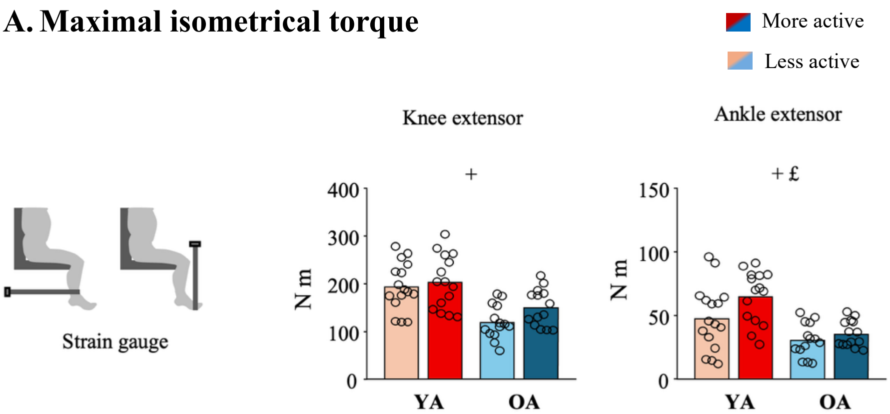


Fig. 3.2A *Maximal isometric torque.* ***A.*** *Maximal Isometric Torque: the schematics illustrate the participant's position and the orientation of the strain gauge fixation, which varies depending on the muscles being assessed. The bar plots represent the maximal isometric torque of knee extensors and ankle extensors for each group. In each subplot, the bars represent the group mean, whereas the open circles indicate individual data. A significant effect of age, physical activity level, and their interaction are indicated with the symbols +,*

*£, and $, respectively. The asterisks indicate post hoc comparisons between groups (*p < 0.05, **p < 0.01, ***p < 0.001, ****p < 0.0001).*

When comparing the mechanical properties of knee and ankle extensors, the stiffness of both rectus femoris and gastrocnemius medialis decreased with age (RF: $F_{1,326} = 101.8$, $p < 0.001$; GM: $F_{1,326} = 5.5$, $p = 0.019$). Instead, the tissue dampening increased for both muscles with age (RF: $F_{1,326} = 199.0$, $p< 0.001$; GM: $F_{1,326} = 169.9$, $p< 0.001$). The physical activity level significantly influenced both stiffness (RF: $F_{1,326} = 76.4$, $p< 0.001$; GM: $F_{1,326} = 21.7$, $p< 0.001$) and tissue dampening (RF: $F_{1,326} = 6.8$, $p= 0.009$; GM: $F_{1,326} = 10.5$, $p= 0.001$;). The active participants exhibited higher muscle stiffness and lower tissue dampening. Regarding stiffnesses, the effect of physical activity level was more pronounced in young adults (interaction: RF: $F_{1,326} = 8.1$, $p= 0.005$; GM: $F_{1,326} = 4.9$, $p= 0.027$) (Fig. 3.2B).

B. Mechanical Properties

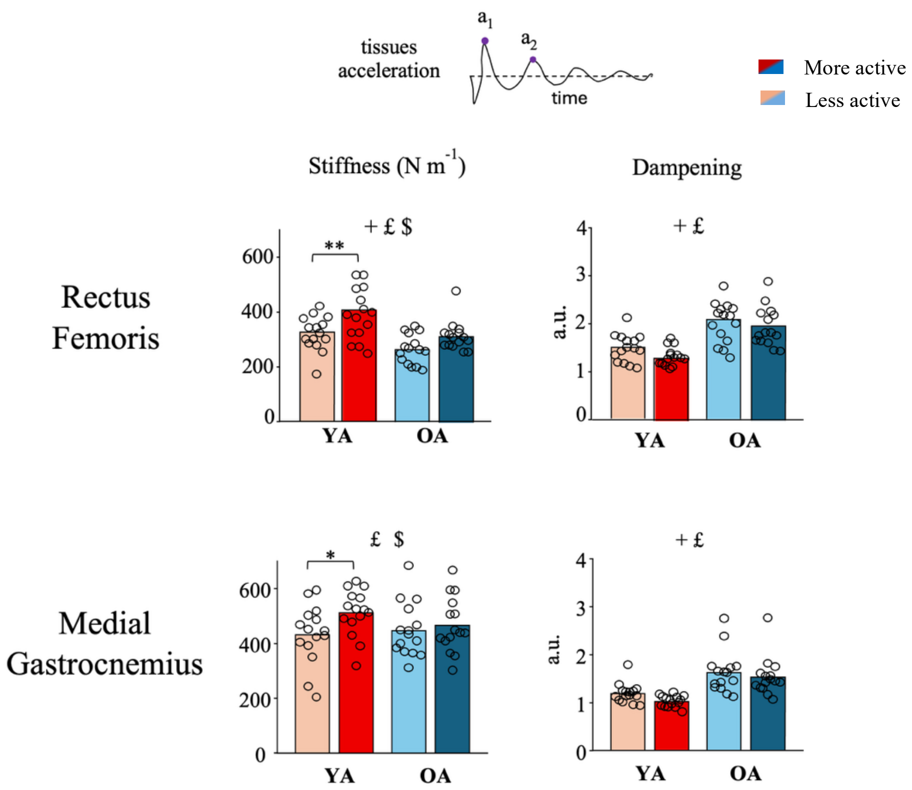


Fig. 3.2B. *Mechanical properties: The stiffness (N/m) and damping (a.u.) of the rectus femoris (RF) and gastrocnemius medialis (GM) muscles were measured using the MyotonPRO device. The upper schematic illustrates the tissue acceleration following the brief impulse applied by the MyotonPRO, which records the subsequent oscillatory response of the tissue. The peak a_1 represents the initial acceleration peak and a_2 represents the decaying oscillations as the tissue returns to its resting state. In each subplot, the bars represent the group mean, whereas the open circles indicate individual data. Other information as in Fig. 3.2A*.

Static balance

The average centre of pressure (CoP) displacement during standing was compared between groups (Fig. 3.2C.). The lateral displacement of the CoP was not influenced by age or physical activity level (Age: $F_{1,54} = 2.4$, $p = 0.121$; physical activity level: $F_{1,54} = 0.1$, $p = 0.710$). Regarding the anteroposterior displacement of the CoP, greater displacement was observed in older adults ($F_{1,54} = 4.0$, $p = 0.050$), but the level of activity did not affect the static balance ($F_{1,54} = 0.56$, $p = 0.457$).

C. Posturography

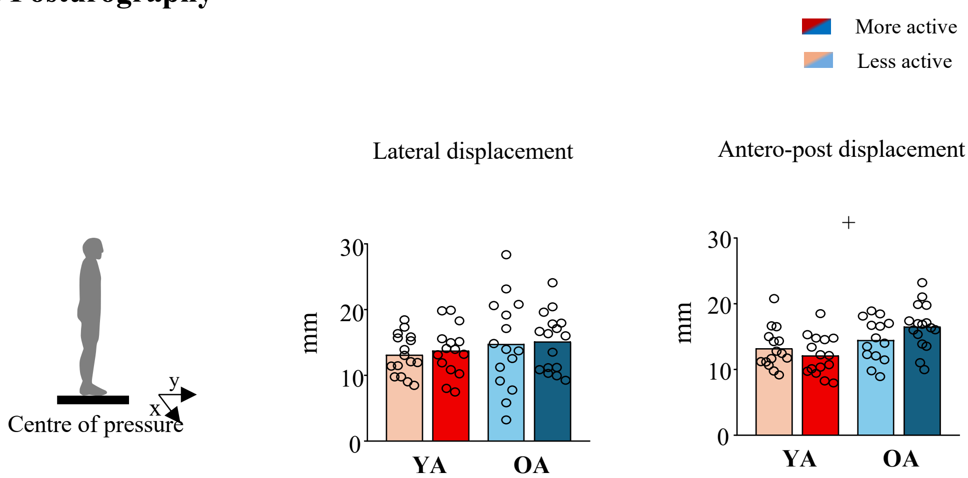


Fig. 3.2C. *Posturography: The CoP displacement was measured in both lateral (X) and anteroposterior (Y) directions. The bar plots indicate the average lateral and anteroposterior displacement in each group of participants. In each subplot, the bars represent the group mean, whereas the open circles indicate individual data. Other information as in Fig. 3.2A.*

Gait mechanics

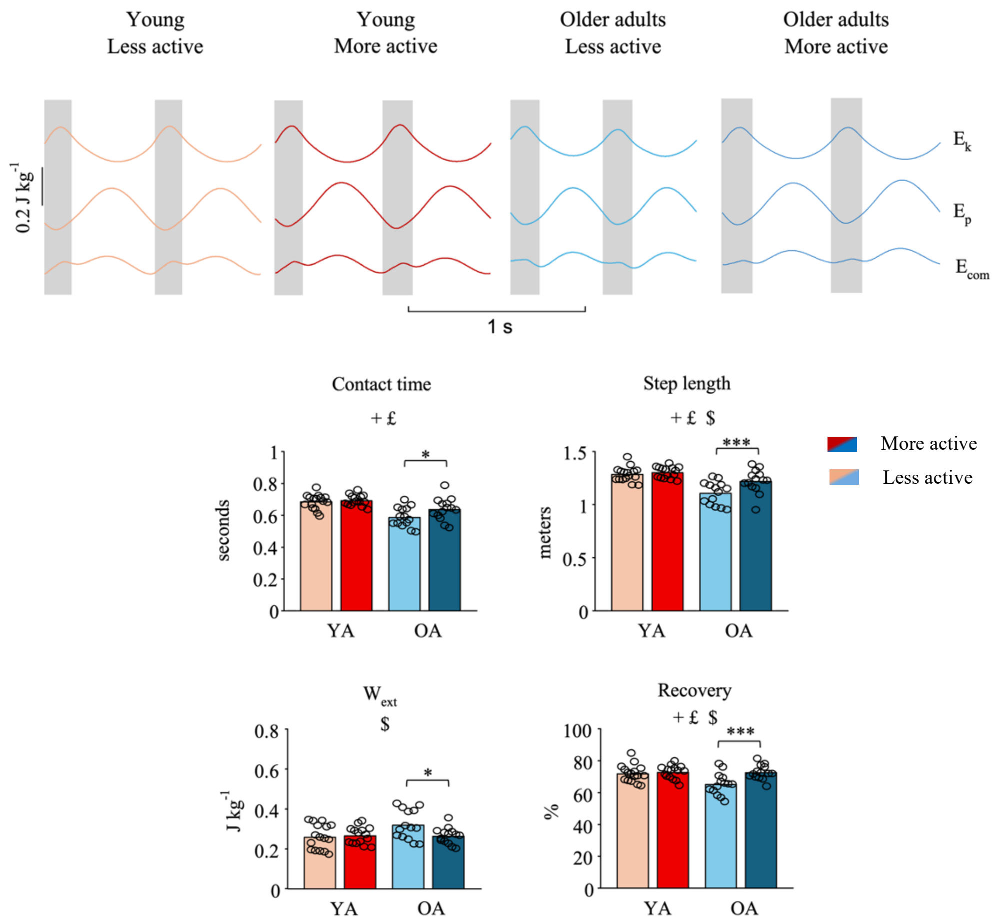
Figure 4.3 illustrates a typical trace of the centre of mass (CoM) energy fluctuation and the time course of the instantaneous recovery within one stride in each group. Older adults walked with shorter steps ($F_{1,55} = 32.8$, $p < 0.001$). Interestingly, the effect of age was greater for less active participants (physical activity level: $F_{1,55} = 8.4$, $p = 0.005$; interaction: $F_{1,55} = 4.4$, $p = 0.040$), with the less active older adults displaying significantly shorter step lengths compared to the more active ones. Similarly, the stance phase was smaller in older adults ($F_{1,55} = 29.6$, $p < 0.0001$), and in less active participants ($F_{1,55} = 4.1$, $p = 0.045$).

The mechanical work required to sustain the motion of the CoM relative to the surroundings (W_{ext}) was not affected by the age ($F_{1,55} = 3.8$, $p = 0.056$) or physical level of the participants ($F_{1,55} = 2.7$, $p = 0.101$). However, an interaction between these two factors was found ($F_{1,55} = 4.4$, $p = 0.040$), showing that in older adults, the mechanical cost of walking was higher in less active individuals compared to the more active ones (post hoc: $p = 0.019$). Such difference in W_{ext} can partially be explained by a reduction of energy transduction between E_k and E_p. Indeed, the *%recovery* was reduced in older adults ($F_{1,55} = 6.3$, $p = 0.014$) in less active participants ($F_{1,55} = 8.8$, $p = 0.004$), with a significant interaction between both fixed factors ($F_{1,55} = 5.4$, $p = 0.023$). In sum, as for W_{ext} no difference was observed between less and more active young adults (post hoc: $p = 0.645$), whereas the less active older adults showed a significantly lower *%recovery* than the more active ones ($p < 0.001$).

The pattern of fluctuation of the CoM power during walking (Fig. 3.3B) typically presents one major peak of negative power ($\dot{W}^-$) and two major peaks of positive power ($\dot{W}_1^+$& $\dot{W}_2^+$) (Cavagna et al., 1976; A. H. Dewolf, Ivanenko, et al., 2019b). The three peaks of power were affected by age ($\dot{W}^-$: $F_{1,55} = 9.5$, $p = 0.003$; $\dot{W}_1^+$: $F_{1,55} = 14.5$, $p < 0.001$; $\dot{W}_2^+$: $F_{1,55} = 14.4$, $p < 0.001$), with greater $\dot{W}^-$ and $\dot{W}_1^+$ and smaller $\dot{W}_2^+$ in older individuals. No significant effect of physical activity level was found (all $p > 0.547$), but an interaction between age and the level of activity was significant for $\dot{W}_1^+$ ($F_{1,55}$

= 4.9, *p*= 0.030). The peak of power tended to be greater in older participants less active compared to more active ones, whereas no differences were observed between young adults less or more active (post hoc: *p*= 0.243).

A



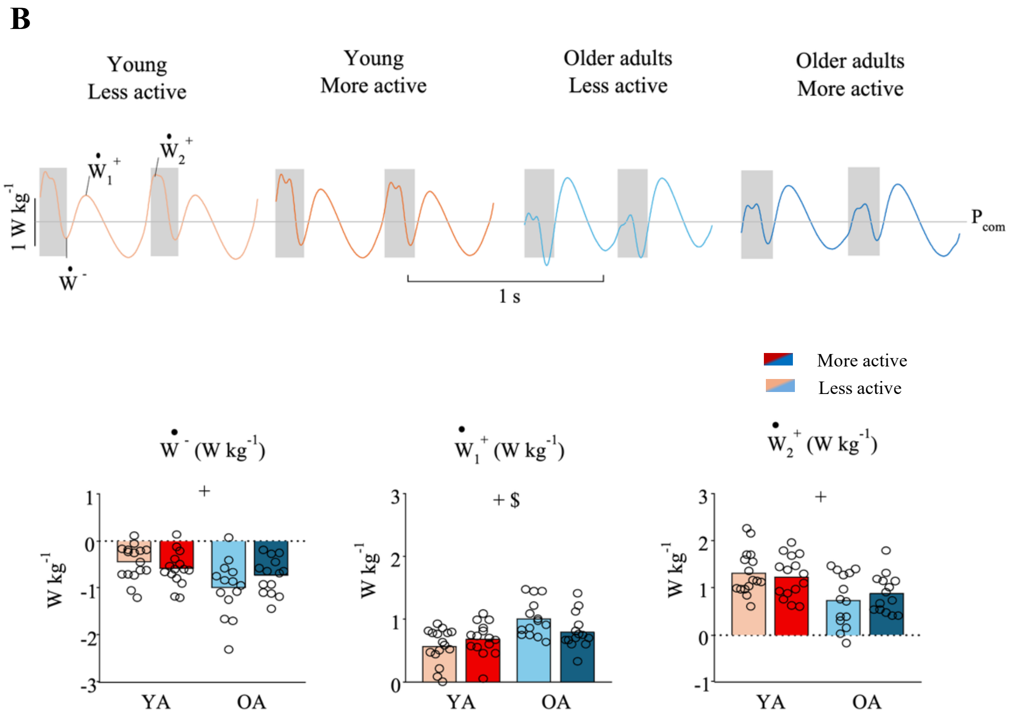
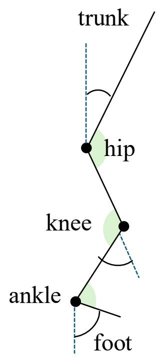


Fig. 3.3. Mechanics of walking. ***A***. *Mean traces of kinetic energy (E_k), potential energy (E_p), and total mechanical energy (E_{com}) versus time across one representative stride for each group of participants. At the bottom, the bar plots indicate from left to right the average stance duration (s), step length (m), external work to sustain the motion of the CoM (J kg^{-1} m^{-1}), and the %recovery.* ***B***. *Mean time curves of the external power of the center of mass (P_{com}) in each group of participants. Each trace displays one major negative peak (W^-, collision) and two major positive peaks, W^+_1 (weight acceptance) and W^+_2 (propulsion), during a stride. The bar plots represent the average value of the three peaks of power. In all panels, the grey shaded areas indicate the double-contact phases. The length of the trace along the x-axis represents the mean stride time for each group. Other information as in Fig. 3.2A.*

Gait kinematics

The average lower-limb joint angles (hip, knee and ankle), as well as two orientation angles (trunk and foot), were measured for each group of participants (Figure 4.4A). In Figure 4.4B, we have highlighted 3 parameters affected by age: the foot angle at touchdown, the mean trunk angle, and peak knee flexion during stance. Indeed, compared to young adults, the foot at touchdown was less dorsiflexed ($F_{1,54} = 4.6$, $p= 0.036$), the trunk was more inclined forward ($F_{1,54} = 28.2$, $p < 0.0001$) and peak knee flexion was greater during stance ($F_{1,54} = 4.7$, $p = 0.033$) in older adults. None of these angles was affected by the level of physical activity (all $p > 0.201$). Nevertheless, an interaction between the effect of age and physical activity level was significant for the foot angle at touchdown ($F_{1,54} = 8.5$, $p = 0.005$). Indeed, the older adults more active had greater ankle dorsiflexion at touchdown compared to less active ones (post hoc: $p = 0.002$), whereas no difference was found between groups of younger adults.

A



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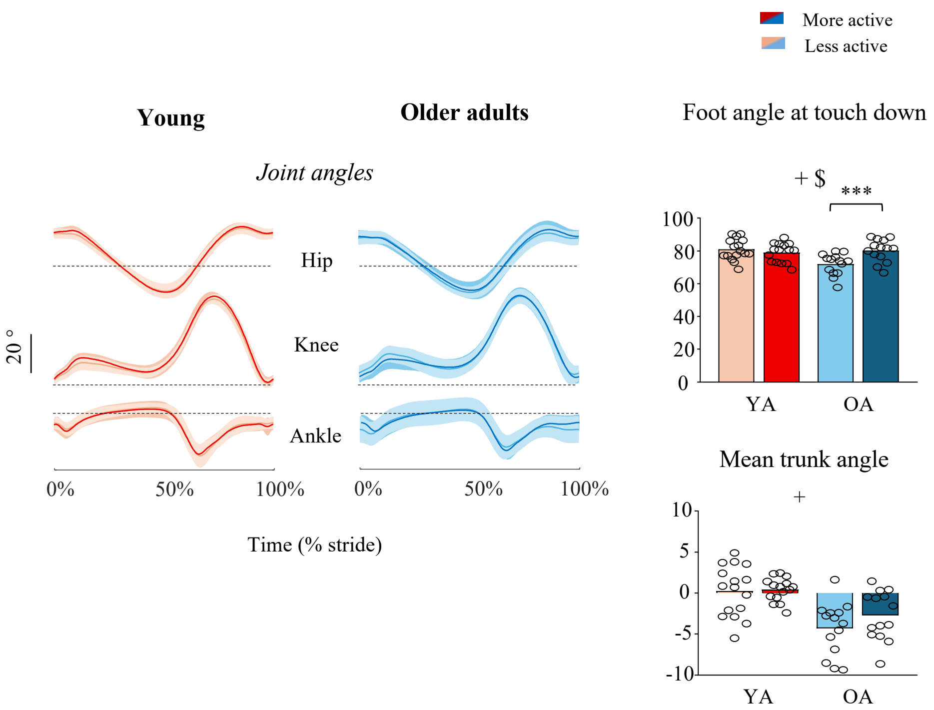


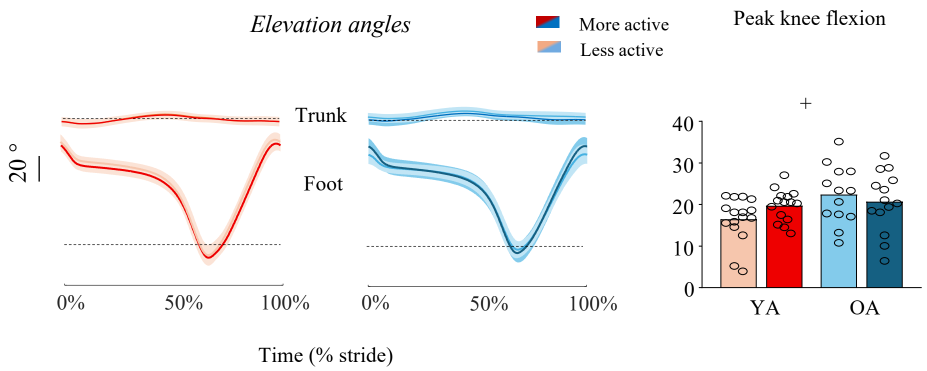


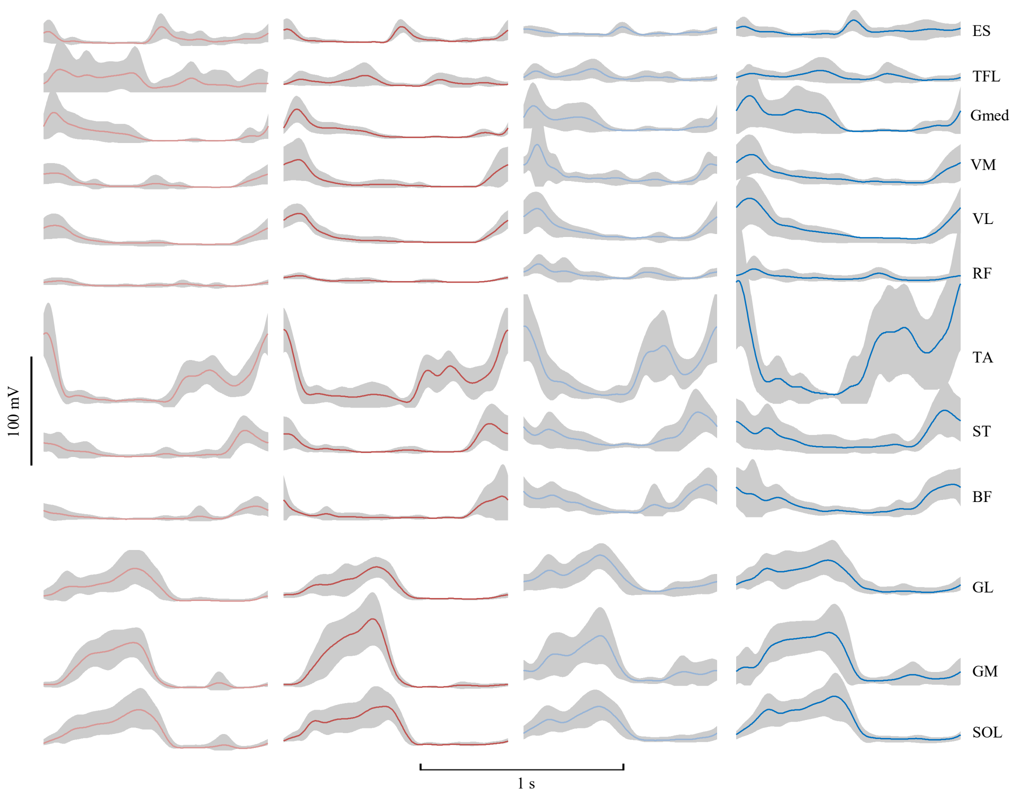
*Fig. 3.4. Kinematics of walking. **A.** The stickman illustrates the different joint angles (hip, knee, and ankle) as well as the two orientation angles (trunk and foot) measured in each participant. The mean traces and standard deviations (shaded areas) of each angle over a stride are presented. **B.** The bar plots present the average foot angle at touch down, the mean trunk angle over a stride, and the peak knee flexion during stance in each group. Other indications as in Fig. 3.2.*

Muscle activity

Figure 4.5 shows the ensemble averages of rectified electromyographic (EMG) envelopes for each group during one walking stride. When comparing the root mean square (RMS) of each EMG, an average greater activation was observed in older adults for all knee extensor and flexor muscles (*vastus medialis, vastus lateralis, rectus femoris, semitendinous, biceps femoris*), and also for *tibialis anterior*, and *gastrocnemius lateralis* (for all F> 6.2 and p< 0.016). Also, an effect of physical activity was observed for some muscles (*vastus lateralis*: $F_{1,53} = 4.8$, $p = 0.032$; *tibialis anterior*: $F_{1,55} = 4.0$, $p = 0.049$; *gastrocnemius medialis*: $F_{1,55} = 6.1$, $p = 0.016$), with a higher activation in active individuals. Moreover, for the *erector spinae*, an interaction between age and physical activity level was found ($F_{1,54} = 5.0$, $p = 0.029$), with lower activation in less active older adults compared to their more active peers (post hoc: p = 0.013). For the *tibialis anterior*, post hoc analysis revealed that older adults in the more active group had higher activation than their less active counterparts ($p = 0.015$), while no differences were found in younger adults ($p = 0.785$).

Regarding the duration of EMG activation, measured based on the full width at half maximum (FWHM) (Fig. 3.5B), longer bursts were observed in older adults for a great majority of muscles ,erector spinae (ES) , tensor fasciae latae (TFL), the gluteus medius (GMED), rectus femoris (RF), tibialis anterior (TA), semitendinosus (ST), long head of biceps femoris (BF), soleus (SOL) (for all F> 6.2 and p<0.015). Instead, no effect of physical activity level (or interaction) was found (for all F< 3.3 and p>0.071). The timing of activation was also affected by age. Indeed, the ES, GMED and BF were activated later in the stride in older adults (ES: $F_{1,54} = 7.1$, $p= 0.010$; GMED: $F_{1,54} = 12.8$, $p= 0.001$; BF: $F_{1,53} = 6.5$, $p= 0.013$), whereas the ankle extensors were activated earlier in stance (gastrocnemius lateralis: $F_{1,55} = 10.4$, $p= 0.002$; GM: $F_{1,55} = 7.7$, p= 0.007; SOL: $F_{1,54} = 5.8$, $p = 0.019$). Again, no effect of physical activity level (or interaction) was found on the timing of activation (for all F< 1 and p>0.319).

A



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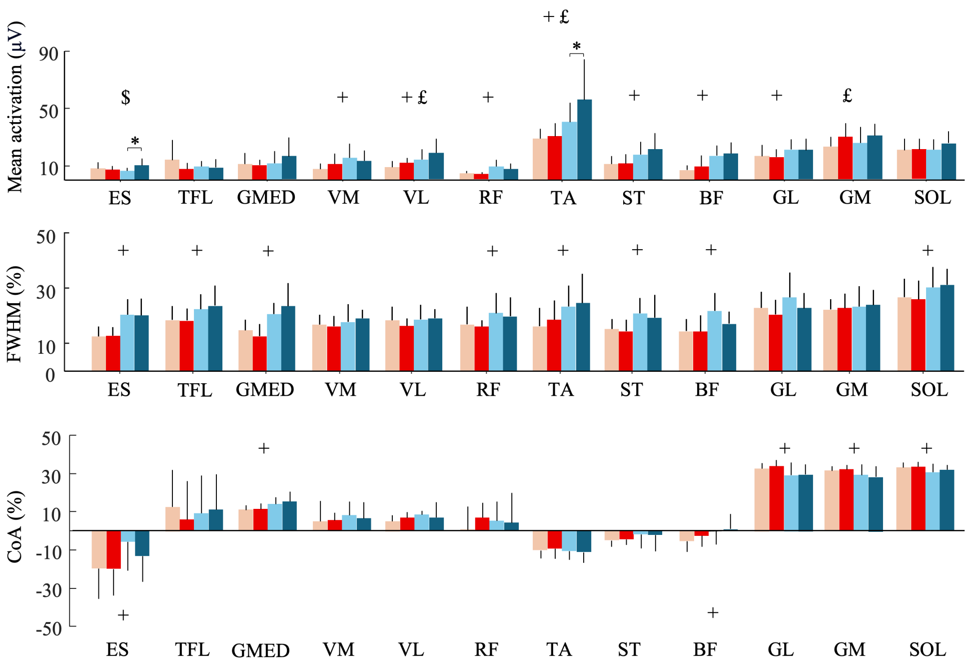

Fig. 3.5. *Electromyography (EMG) activation patterns.* ***A.*** *Ensemble-averaged EMG curves of the erector spinae (ES), tensor fasciae latae (TFL), gluteus medius (Gmed), vastus medialis (VM), vastus lateralis (VL), rectus femoris (RF), tibialis anterior (TA), semitendinosus (ST), biceps femoris (BF), gastrocnemius lateralis (GL), gastrocnemius medialis (GM), and soleus (SOL). The colored lines represent the ensemble-averaged EMG traces for each group, while the shaded grey areas indicate the standard deviation.* ***B.*** *The bar plots represent, from top to bottom, the root mean square (in µV), the full-width half maximum (FWHM, in % of stride), and the center of activation (CoA, in % of stride) for each muscle and each group. The bars correspond to the mean values, with vertical lines above each bar representing one standard deviation. Other indications as in Fig. 3.2.*

Spinal maps

The Figure 4.6A presents the EMG signals mapped onto the estimated rostro-caudal location of the motor neuron (MN) pools in the spinal cord. The timing of activation of the lumbar and sacral motor pools, estimated using the centre of activity (CoA), was different between young and older participants (lumbar: $F_{1,55} = 4.8$, $p = 0.032$; sacral: $F_{1,55} = 22.0$, $p < 0.001$). Indeed, the lumbar CoA occurred later whereas the sacral CoA occurred earlier. As a result, the delta between lumbar and sacral activations was reduced in older adults ($F_{1,55}$ = 23.4, p < 0.001). While no effect of physical activity was observed on individual muscle activation patterns (Fig. 3.6), an interaction was found between age and level of physical activity for the lumbar CoA ($F_{1,55}$ = 15.6, p< 0.001), with less active older participants displaying a later activation of the lumbar motor pools than more active elders (post hoc: p < 0.001), without difference observed between less and more active young participants (post hoc: p = 0.082). In turn, the delta between lumbar and sacral CoA was also smaller for less active older adults than more active ones (interaction: $F_{1,55}$ = 8.8, p= 0.003; post hoc: p = 0.003).

The burst durations of the spinal motor output were also affected by age, in both lumbar ($F_{1,55}$ = 12.2, p= 0.001) and sacral segments ($F_{1,55}$ = 7.3, p= 0.008). Most likely due to longer bursts, the co-activation level (CI) between the lumbar and sacral motor pools was greater in older adults ($F_{1,55}$ = 21.4, p < 0.001). While no effect of physical activity level on burst duration was found in both spinal levels (p > 0.05), the co-activation was higher in less active individuals ($F_{1,55}$ = 4.7, p = 0.034), with a higher CI in less active older adults than more active ones (post hoc: p = 0.029).

The mean activity of both the lumbar and sacral segments was higher in older adults ($F_{1,55}$ = 26.6, p < 0.001; $F_{1,55}$ = 21.8, p < 0.001), but the ratio between lumbar and sacral activation was similar in both age groups ($F_{1,55}$ = 3.9, p= 0.053). No significant effect was found for the physical activity level in the mean activation of the spinal segments (lumbar: $F_{1,55}$ = 0.7, p = 0.616; sacral: $F_{1,55}$ = 0.7, p = 0.384) or in the ratio between lumbar and sacral activation ($F_{1,55}$ = 0.002, p = 0.966).

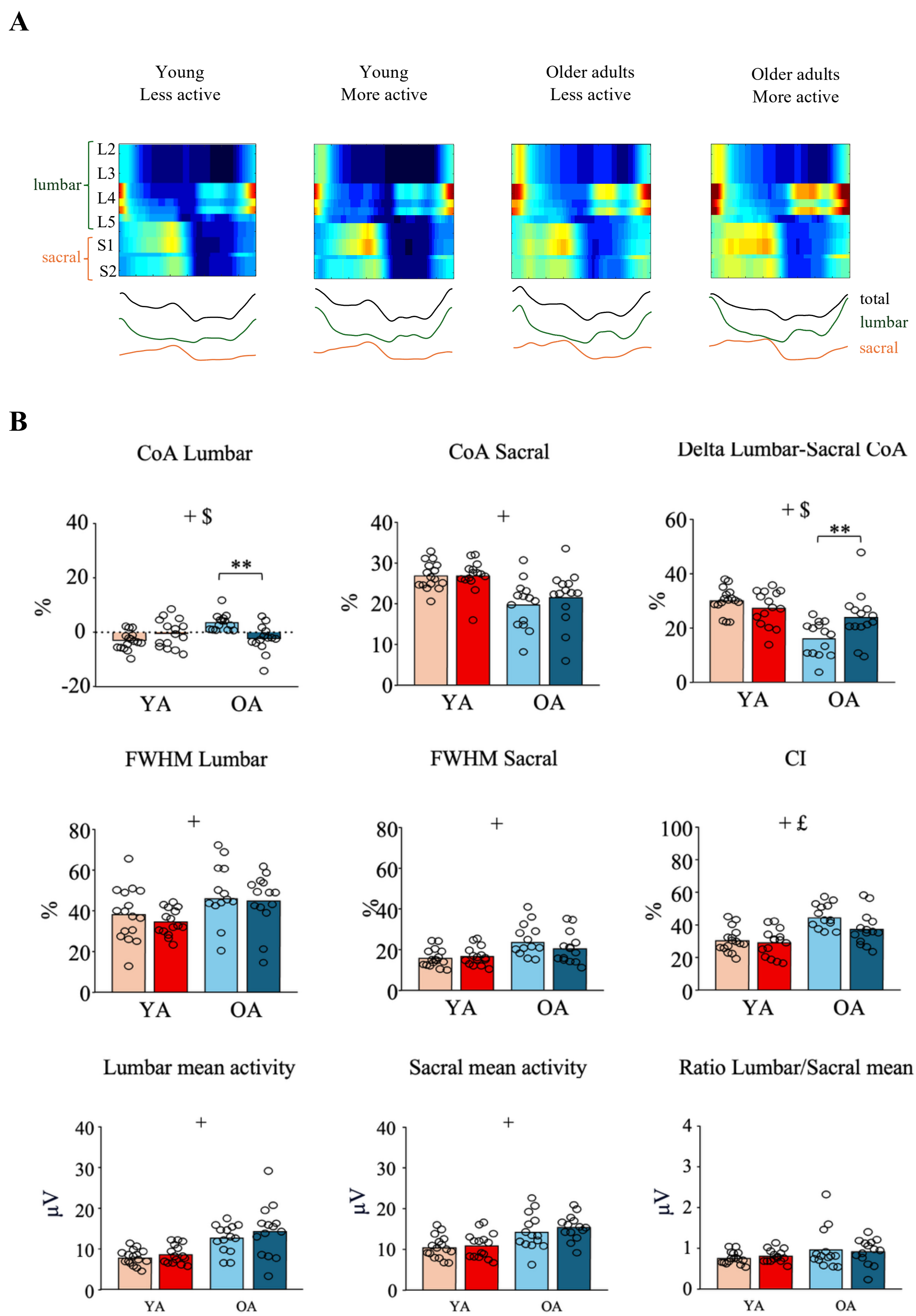
A
Young
Less active
Young
More active
Older adults
Less active
Older adults
More active
lumbar
sacral
L2
L3
L4
L5
S1
S2
total
lumbar
sacral
B
CoA Lumbar
CoA Sacral
Delta Lumbar-Sacral CoA
FWHM Lumbar
FWHM Sacral
CI
Lumbar mean activity
Sacral mean activity
Ratio Lumbar/Sacral mean
+ $
+
+ £
**
%
μV
YA
OA

Fig. 3.6. *Spinal motor outputs.* ***A.*** *Unilateral spatiotemporal spinal motor outputs computed from ensemble-averaged EMGs for all participants of each group. The total, lumbar, and sacral activation levels are also depicted as traces below the heat maps for each group, relative to the gait cycle.* ***B****. Bar plots show the center of activation (CoA, in % of stride) for the lumbar and sacral segments and the delta between lumbar and sacral CoA. Also, the full-width half maximum (FWHM, in % of stride) for both spinal regions, and the coactivation index (CI, %) between lumbar and sacral activations are presented. Finally, the mean activation levels (in µV) of lumbar and sacral outputs, along with the ratio between lumbar and sacral mean activation are displayed. Other indications as in Fig. 3.2.*

Discussion

In this study, we provide new insights into how a higher level of physical activity can influence the age-related changes observed during walking. Gait in older adults is often characterized by a shorter step length, greater metabolic cost, and distal-to-proximal joint effort redistribution. Our results show that the less active older subjects significantly decrease their step length, increase the power at collision via proximal joints and increase their mechanical work during walking (by about ~20% relative to young adults; Fig. 3.3A) relative to more active individuals. While we confirm our hypothesis regarding the impact of aging on muscle mechanical properties and maximal strength, relatively few changes are observed between the most active individuals and the more sedentary ones. However, a modification in spinal motor output was observed in less active individuals, showing an accentuation of the previously documented effect of aging.

The metabolic cost of walking is an important variable of daily life that has been studied extensively. In systematic reviews, Das Gupta *et al.* (2019) and Aboutorabi *et al.* (2016) showed that the average net cost of walking was ~20% higher in older adults, associated with a loss of independence and an increased risk of morbidity and mortality (Cooper et al., 2010, 2011; Guralnik et al., 1995; Studenski et al., 2011). Previous studies have found that mechanical work does not explain the increase in the metabolic cost of elderly walking since no significant changes have been observed (Mian et al., 2006; Ortega & Farley, 2007). Considering only the effect of age, we did not observe any significant change in mechanical work between younger and older individuals (Fig. 3.3A). However, considering the level of physical activity, we observed a 20% increase in mechanical work in less active older adults. This change can partly be linked to a decrease in the pendular mechanism of walking in less active older adults (Fig. 3.3A).

In a previous publication, we reported a similar reduction of energy recovery in older adults unable to accelerate the centre of mass forward and upward before double stance because of a lack of trailing leg push-off at the end of stance (Dewolf, Sylos-Labini, Cappellini, Zhvansky, et al., 2021a; Dewolf et al., 2022). Consequently, the centre of mass accelerates after the foot contact of the leading limb (Meurisse et al., 2019b). Similarly, our results showed that the phase of negative power ($\dot{W}^{-}$) occurring after collision with the ground, followed by a first phase of positive power ($\dot{W}_1^{+}$) were higher in older adults, and specifically in less active ones (Fig. 3.3B). Both contribute to the vertical redirection of the centre of mass after foot contact (FC), via greater eccentric energy at the knee extensors of the leading limb (Lay et al., 2006; Monaco & Micera, 2012) and were designated as the collision phase (A. H. Dewolf, Ivanenko, et al., 2019b). Also, a reduction of the second phase of positive power ($\dot{W}_2^{+}$) was reduced in older adults, suggesting a reduction of ankle propulsion in late stance (DeVita & Hortobagyi, 2000; Silder et al., 2008). Such distal-to-proximal redistribution with age has garnered considerable scientific attention for nearly 50 years (Delabastita et al., 2021; DeVita & Hortobagyi, 2000; Franz, 2016; Franz & Kram, 2013; Gueugnon et al., 2019a; Kerrigan et al., 2000; McGibbon, 2003; Winter, Patla, et al., 1990), most often consider as the hallmark biomechanical ageing features of gait.

While one might expect to observe a change in propulsive force in more active individuals, the age-related reduction of propulsion was similar between groups. Similarly, it has been shown that resistance training consistently improves muscle strength and mitigates sarcopenia but fails to directly translate to improved propulsive power generation in walking (Beijersbergen et al., 2013). Instead, the impact of physical activity level was mainly observed during the collision phases (Fig. 3.3B). Such increase has been associated with deviations from the specific adaptations of the locomotor apparatus to bipedal gait, including the erect posture of the trunk (Núñez-Lisboa et al., 2024), straight leg during stance (Cavagna & Kaneko, 1977;

Full & Koditschek, 1999), and heel-to-toe rolling pattern of the foot (Mesquita et al., 2023; Usherwood et al., 2012). Our results showed that less active older adults reduced the heel-to-toe rolling pattern, by decreasing dorsal flexion at foot contact (Fig. 3.4B). Also, less active individuals tend to increase knee flexion during stance and to increase the average trunk flexion (Fig. 3.4). All these modifications of kinematic strategy have been related to a greater impact under the leading limb during step-to-step transition (Mesquita et al., 2023; Núñez-Lisboa et al., 2024), as in older adults. These alterations could stem from age-related declines in neuromuscular function, and the nervous system may adapt by modifying motor strategies to enhance stability and reduce the risk of falls.

Neural factors are likely to contribute as well to age-related motor impairment (Nùñez-Lisboa et al., 2023). For example, changes in control strategy (Hortobágyi et al., 2011; Peterson & Martin, 2010a; VanSwearingen & Studenski, 2014) have been reported in older adults. Indeed, compared to young adults, older adults exhibited greater activation of lower-limb muscles during walking (Schmitz et al., 2009), and higher co-activation of antagonist muscles across the ankle and knee potentially contributing to joint stiffening (Marques et al., 2013; Peterson & Martin, 2010a). In our results, we observed a greater activation with age in *tibial anterioris* and gastrocnemius lateralis, as well as in knee extensors and flexors (Fig. 3.5). Also, the timing of muscle activation changed with age, being longer for both ankle flexor (TA) and extensor (SOL) muscles, and knee flexors (ST, BF) and extensor (RF), potentially increasing co-activation.

The greater muscle activation could be related to the decline in muscle quality. Aging muscles weaken (Fig. 3.2B), which results in an increase in activation and thus in more energy consumed per amount of force produced (Delmonico et al., 2009; Goodpaster et al., 2006). In addition, following the principle of motor unit recruitment, motor unit recruitment progresses from slow-twitch to metabolically more expensive fast-twitch fibres at higher activations (Bhargava et al., 2004). While the effect of age is in accordance with previous published results, little to no modifications were observed in older adults based on their physical activity level, except for EMG amplitude.

Indeed, some muscles (TA, vastus lateralis, GM and ES) were more active in less active individuals.

Interestingly, insights may be gained by looking into the recruitment and organization of motoneurons during locomotion. The spinal maps, assessed by mapping muscle activity onto the rostrocaudal location of motoneurons, have already revealed specific characteristics of normal aging (Avaltroni et al., 2024; Dewolf et al., 2021a, 2021b; Monaco et al., 2010; Nùñez-Lisboa et al., 2023). As compared to young adults, the motoneurons activation is significantly wider in older adults and the delta between lumbar and sacral activation is reduced. Indeed, the lumbar and sacral activity centres are closer together (Fig. 3.6), inducing a higher co-activation. Such modification has been related to the lack of late push-off prior to toe-off from the trailing leg (Avaltroni et al., 2024; Dewolf et al., 2021b; Nùñez-Lisboa et al., 2023). As for kinematics, it's interesting to note that we observed a deviation from the two features of motoneuron activation that have been associated with the most energy-efficient bipedal gait: 1) distinct bursts of activity in the lumbar and sacral segments and 2) short burst durations relative to gait cycle duration (Avaltroni et al., 2024).

While many age-related changes in gait patterns have been related to physiological deficits in peripheral system function (Horak, 2006), our results suggest that changes in the way the central nervous system coordinates the movements also contribute. In addition, our results show that age-related changes are amplified in older individuals with lower levels of physical activity. For example, the CoA of lumbar motor pools occurred after touch down in less active older adults, as compared to the other groups for whom the activity on average occurred before touch down. It potentially reflects a lack of anticipation before the collision with the ground, as reported earlier in older adults with a higher risk of falls (Meurisse et al., 2019b). Also, greater co-activation is observed between the activity of the two motoneuron pools in older adults less active, compared to active ones (Fig. 3.6B). This co-activation reflects the widening of spinal motor output, which has been proposed as a conservative strategy aimed at maintaining gait stability (Martino et al., 2015; Santuz et al., 2020). This seems to support our hypothesis that engaging in physical activity helps slow down the degradation of the neuromuscular system. Indeed, it's plausible that regular physical activity may mediate the age-related changes in the central nervous system, and in turn affects movement control strategy (Santuz et al., 2020; Y. Wang et al., 2019).

The modification of control strategy is also coupled with other intrinsic factors, such as weaker muscles and softer muscle-tendon units in older adults (Monaco & Micera, 2012). Indeed, both knee and ankle extensor maximal isometric torque was reduced with age (Fig. 3.2B), and individuals with less physical activity displayed reduced ankle extensor maximal forces. Reduced physical activity typically results in skeletal muscle atrophy and an associated reduction of maximal voluntary force (Venturelli et al., 2015). Numerous alterations in the physiological properties of the locomotor apparatus have been observed with aging (Pardes et al., 2017). Using a predictive neuromechanical model to investigate the physiological origins of age-related gait changes, Song and Geyer (2018) showed that the loss of muscle strength and muscle contraction speed dominantly contribute to the reduced walking economy (S. Song & Geyer, 2018). The smaller ankle maximal force of less active individuals may exacerbate age differences and induce a greater redistribution of lower extremity joint moment and power compared with more active ones. Such results could explain why vigorous exercises, such as power training or running, prevents the age-related deterioration of muscle (Beijersbergen, Granacher, Gäbler, Devita, et al., 2017b) and, make everyday activities easier (Beck et al., 2016).

In addition to strength, age-related changes in the mechanical and morphological properties of lower-limb muscle-tendon units have been reported (Karamanidis & Arampatzis, 2005). Our results also showed that the muscle-tendon stiffness, measured via myotonmetry, was affected by age: the stiffness of rectus femoris and gastrocnemius lateralis was reduced and the dampening (oscillation attenuation) was increased in older adults (Fig. 3.2A). It has been suggested that age-related changes in neuromuscular activity (higher coactivation) reflect a strategy of stiffening the limb during stance likely to compensate the reduced muscle-tendon stiffness, to utilize tendon elasticity effectively (Gollhofer et al., 1992). Indeed, it is suggested that the muscle fascicle behaviours can be related to the motor strategy of older adults to stiffen joints and stabilize motor output in an effort to compensate for reduced muscle strength and for increased tendon compliance and joint laxity (Hortobágyi & DeVita, 2006). Interestingly, a higher level of physical activity in older adults reduced the differences between young and older adults, suggesting a positive impact of the level of activity (Fig. 3.2A). In addition to the change in strength, this effect of physical activity may contribute to the modification of motor strategy observed between less and more active older individuals, in particular the reduction of co-activation between lumbar and sacral segments (Fig. 3.6).

When comparing older adults according to fall history (non-fallers versus fallers groups), modification of gait and neuromuscular system between groups was similar to the one observed in the present study between less and more active older adults. Indeed, fallers showed shorter stride length (Mortaza et al., 2014), increased cost of walking, higher co-activation of lower-limb muscles (Marques et al., 2013) and modified motor control strategy (J. L. Allen & Franz, 2018). In addition, muscle stiffness of the *gastrocnemius medialis* was significantly lower in the fallers group (Baş et al., 2023) and older adult fallers had 28% lower knee extensor strength (Marques et al., 2013). Here, we did observe a more variable static balance between young and older participants (Fig. 3.2C), but no effect of physical activity level. However, given that fallers are presumably less active due to their mobility limitations and fear of falling, it is plausible that their reduced activity levels directly influence these neuromuscular deficits. These findings strengthen our results by highlighting that physical activity plays a crucial role in mitigating age-related changes in gait and neuromuscular function. This reinforces the idea that maintaining or increasing physical activity in older adults can help preserve gait mechanics, and muscle function, and potentially reduce the risk of falls.

The primary limitation of this study is that the level of physical activity was assessed using a self-reported questionnaire, which is inherently less accurate than objective measures such as VO_2 max (Tangen et al., 2022), daily step count (Kraus et al., 2019), or accelerometer-based activity tracking (Metcalf et al., 2018). These more precise methods could provide a deeper understanding of the relationship between physical activity and gait parameters. However, we believe that our approach also serves as a strength of the study. Despite using a simple classification of activity levels, we observed significant modifications in gait and neuromuscular function related to physical activity. In addition, the individuals classified as 'less active' performed an average of 1473.6 METs min/week, exceeding the World Health Organization recommendations (Bull et al., 2020). This reinforces the robustness of our findings and supports the main conclusion of the paper, suggesting that even small increases in physical activity could have meaningful effects on gait. For example, transitioning from a lower activity level to a more active group, with just a few additional hours of sports per week, could result in significant improvements in gait mechanics and mobility for less active older adults.

Another limitation of the study is the use of a treadmill. Walking on a treadmill may potentially reduce gait variability as compared to over-ground gait (Lee & Hidler, 2008). However, treadmill walking has been commonly used to investigate age-related modification of gait (Dewolf, Sylos-Labini, Cappellini, Ivanenko, et al., 2021; Dewolf et al., 2022; Meurisse et al., 2019b), and spatiotemporal, kinematic, kinetic, muscle activity, and muscle-tendon outcome measures are largely comparable between motorized treadmill and overground locomotion (Van Hooren et al., 2020).

In conclusion, the recognition that the mechanical cost of walking is not purely age-dependent, but rather a consequence of reduced physical activity and subsequent skeletal muscle changes, underscores the vital importance of maintaining physical activity across the lifespan to mitigate the effects of aging on mobility. Indeed, our findings demonstrate that the level of physical activity significantly influences age-related changes in gait and neuromuscular function. While some of these changes are an inherent part of the aging process (Pearson et al., 2002), a substantial portion can be attributed to the decline in physical activity levels with advancing age (G. R. Hunter et al., 2000). Our results underline the critical role of regular physical activity in preserving gait mechanics and muscle function in aging. Therefore, incorporating structured training programs that enhance physical activity levels could be an effective strategy to improve walking performance, reduce walking costs, and delay mobility impairments in older adults (Valenti et al., 2016). Such interventions (Malatesta et al., 2010) may ultimately contribute to reducing the risk of falls and promoting greater independence as people age.

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Chapter IV

Acute Spinal Neuromodulation Alters Trunk Posture and Sacral Motor Output in Older Adults during Gait

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List of Abbreviations

BW:	Body weight
CI:	Co-activation index
CoA:	Center of activity
CoM:	Center of mass
COP:	Center of pressure
EMG:	Electromyography
ES:	Erector spinae
F_{back}:	Backward ground reaction force peak
F_{front}:	Forward ground reaction force peak
Fv:	Vertical ground reaction force
FWHM:	Full width at half maximum
GM:	Gastrocnemius medialis
GL:	Gastrocnemius lateralis
Gmed:	Gluteus medius
GRF:	Ground reaction force
MN:	Motoneuron
OA:	Older adults
RF:	Rectus femoris
RMS:	Root mean square
ROM:	Range of motion
S1–S2:	Sacral spinal segments
SMO:	Spinal motor output
SOL:	Soleus
SPM:	Statistical parametric mapping
ST:	Semitendinosus

TA:	Tibialis anterior
TFL:	Tensor fasciae latae
tsDCS:	Trans-spinal direct current stimulation
VL:	Vastus lateralis
VM:	Vastus medialis
$V_{v,min}$:	Minimum vertical velocity of the CoM
YA:	Young adults

Abstract

The objective of this study was to investigate whether a single 20-minute session of trans-spinal direct current stimulation (tsDCS) applied over the sacral region can modulate the spinal motor output and improve gait neuromuscular control in young and older adults. Forty participants (20 young and 20 older adults) completed pre- and post-intervention treadmill walking trials. Participants were randomized to receive either active tsDCS or sham stimulation. Spinal motor output was assessed by mapping multi-muscle EMG activity onto the spinal cord. Ground reaction forces and joint kinematics were analyzed to evaluate gait. No significant changes were observed in spatiotemporal gait parameters or overall gait dynamics. However, in older adults, a single tsDCS session reduced trunk flexion and induced changes in the spinal motor output, specifically within sacral segments. Sham stimulation had minimal effects. These neuromuscular adjustments did not translate into significant kinetic changes. In conclusion, the tsDCS over the sacral region modulated trunk posture and spinal motor output in older adults. These preliminary effects suggest segment-specific neural responsiveness to spinal neuromodulation. This study provides initial evidence that tsDCS can influence postural control and spinal segmental output during gait in older adults, supporting its potential as a non-invasive intervention to address age-related neuromuscular decline.

Introduction

The ability to walk safely and independently is a key determinant of autonomy and quality of life in older adults (Freiberger et al., 2020). As life expectancy increases, preserving mobility becomes a central challenge in aging populations due to its close link with functional capacity, fall risk, and overall health outcomes (Fielding et al., 2011; Geirsdottir et al., 2022; Studenski, 2011). With aging, there is a well-documented redistribution of joint mechanics from distal to proximal segments, where older adults rely less on the ankle and more on the hip to generate forward progression (Delabastita et al., 2021; Franz, 2016; Waanders et al., 2019). This shift also encompasses increased anterior trunk flexion, another hallmark of aging gait (Honda et al., 2023), which has been associated with mechanical alterations, specifically in double contact phase (Núñez-Lisboa et al., 2024). This postural strategy may compensate for reduced plantar flexor function but has also been linked to increases in both mechanical and metabolic energy cost during walking (Delabastita et al., 2021; Dewolf et al., 2022).

Although biomechanical changes clearly influence gait with aging, numerous studies have shown that the age-related alterations in walking patterns cannot be solely attributed to a decline in propulsive power generation (Avaltroni et al., 2024; Boyer et al., 2017; Hortobagyi et al., 2011; Hortobágyi et al., 2009; Monaco et al., 2010; Peterson & Martin, 2010b). Neural factors also appear to play a key role. In particular, consistent age-related differences in muscle activation patterns have been observed across a wide range of walking tasks in older adults—including walking at various speeds, walking backward, walking on slopes, and negotiating stairs—despite differing biomechanical demands and propulsion requirements (Dewolf et al., 2021a, 2021b). By mapping multi-muscle EMG activity onto the spinal cord in correspondence with the approximate rostro-caudal locations of motor neuron pools, it is possible to infer segmental spinal motor output during locomotion (Cappellini et al., 2010; Dewolf et al., 2019a; Ivanenko et al., 2008, 2013; La Scaleia et al., 2014; Yokoyama et al., 2017). Dewolf et al. (2021a & 2021b) demonstrated that older adults exhibit wider and earlier activation of muscles innervated by sacral segments compared to younger individuals. These findings suggest a systematic shift in the spatiotemporal organization of spinal motor output with age. Several strategies have been proposed to reverse age-related gait deficits, including physical exercise for its neuromuscular benefits (M. D. Allen et al., 2021; Hepple & Rice, 2016). However, its impact on spinal motor control remains limited (Beijersbergen et al., 2013; Boyer et al., 2012), prompting interest in alternative neuromodulatory approaches.

In recent years, there has been growing interest in understanding the functional organization of the spinal locomotor output in gait pathologies (Coscia et al., 2015; Grasso, 2004; Martino et al., 2018). With it, selective stimulation of spinal motor pools has been proposed as a promising tool to restore the functioning of the spinal pattern generation circuitry in humans (Angeli et al., 2018; Gerasimenko et al., 2015; Solopova et al., 2017; Wagner et al., 2018). For example, trans-spinal direct current stimulation (tsDCS) is a neuromodulatory approach that target Ca2+ conductance to augment motoneuron activity in the spinal cord (Song & Martin, 2017). After a spinal cord injury, the tsDCS can significantly alter muscle activity during locomotion by modulating spinal motor neuron excitability and spinal circuitry (Hubli et al., 2013). Also on healthy subject, one session of tsDCS enhance body explosive power production (Berry et al., 2017). During walking, it has been recently showed that tsDCS affects the joint kinematics and promotes a more stable coordination (Skiadopoulos & Knikou, 2024).

In this context, tsDCS has emerged as a promising non-invasive technique to modulate spinal excitability and enhance sensorimotor integration (Baczyk, M., 2019; Jankowska, 2017; W. Song & Martin, 2017) during gait. Despite its potential, tsDCS has yet to be explored in the context of aging. Given that older adults exhibit increased amplitude and earlier activation in sacral spinal segments, possibly reflecting greater central integration demands during gait (Avaltroni et al., 2024; Dewolf et al., 2021b; Monaco et al., 2010), we hypothesize that applying anodal tsDCS over the sacral region may modulate this output pattern and improve gait coordination in aging individuals. Given its integrative role, modulating sacral output may also influence gait kinematics and kinetics. To test this, we examined changes in spinal motor output and gait parameters before and after a single session of tsDCS in both younger and older adults.

Methods

Participants

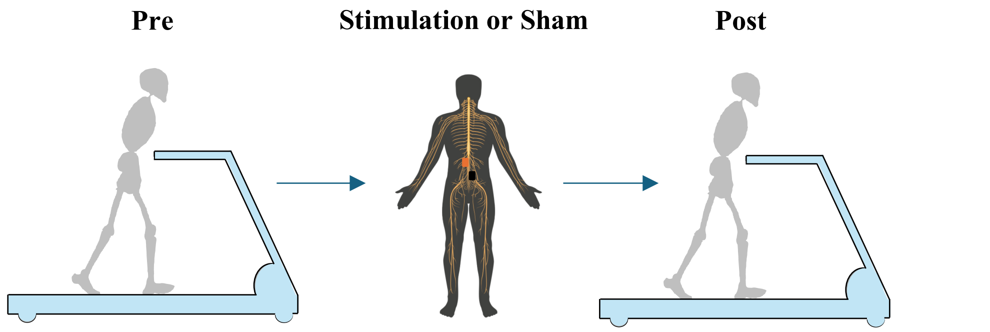
Twenty younger adults (27.2 ± 4.1 y.o.) and twenty older adults (71.5 ± 4.5 y.o.) were included in the study (see Table 4.1 for individual characteristics). Due to time constraints, one older adult voluntarily terminated their participation during the recording session, and was therefore not included in the results. The number of subjects was determined based on the results of tsDCS in healthy young adults(Awosika et al., 2019). Thea priori power analysis using G*Power 3.1.9.7 (effect size = 2.21, $\alpha = 0.05$, power = 0.95) indicated that a minimum of 8 participants per condition was required. However, to ensure adequate power and account for a potential dropout, we included 10 participants per group (sham and stimulation) within each age category. Participants were randomly assigned to either the sham or stimulation group. The inclusion criteria were the following: ability to walk a kilometer, no locomotor system injury complaints, and no previous history of neurological disorders. All participants provided informed consent, and the procedures used for this study were approved by the ethics committee of the Université Catholique de Louvain (Belgian Registration Number: B403201524765) and adhered to the Declaration of Helsinki.

Experimental protocol

The participants were asked to walk wearing their shoes on an instrumented treadmill at 1.11 m s^{-1} (4 km h^{-1}). The selected walking speed was similar to the one used in our previous studies on older adults (Dewolf, Meurisse, et al., 2019; Dewolf, Sylos-Labini, Cappellini, Ivanenko, et al., 2021; Dewolf, Sylos-Labini, Cappellini, Zhvansky, et al., 2021a; Dewolf et al., 2022) and was reported as the comfortable and economical average walking speed based on the net energy consumption (Cavagna & Kaneko, 1977). Before starting data collection of walking, participants familiarized themselves with treadmill walking through multiple trials with verbal feedback from the experimenter. After the familiarization period, we gradually ramped up the treadmill to 4 km h^{-1} and then waited at least 30 seconds before starting the kinematics, GRF and electromyographic data acquisition (Fig. 3.1).

Concerning the stimulation procedure, tsDCS was delivered at 2.5 mA for 20 minutes using a battery-driven, programmable direct current stimulator (Soterix Medical, USA), connected to two surface electrodes (10 × 5 cm) (Awosika et al., 2019). The anode (or sham electrode) was positioned over the spinous process of T11, aligned rostro-caudally with the spinal axis to target segmental spinal reflex excitability. The reference (cathodal) electrode was placed over the lumbosacral region (Fava De Lima et al., 2022). The electrode orientation was intended to promote rostro-caudal current flow, consistent with spinal activation patterns during locomotion. The total charge density delivered was approximately 85.7 mC/cm², remaining well within established safety limits. The same treadmill walking task was repeated immediately after the stimulation session. Participants completed a baseline trial (pre) before tsDCS application, and a second trial (post) under identical conditions. Considering both the pre- and post-stimulation trials, an average of 56.2 ± 5.8 strides were analyzed per participant.

A



B

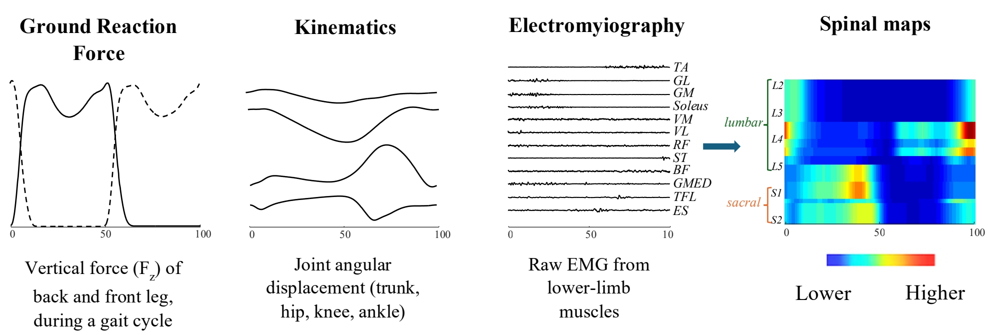


Fig. 4.1. *Experimental design and data acquisition. (**A**) Participants performed treadmill walking before (Pre) and after (Post) a 20-minute trans-spinal direct current stimulation (tsDCS) or sham session. The stimulation targeted the thoracolumbar spinal cord through surface electrodes positioned over T11 (anode) and the lumbosacral region (cathode). (**B**) Biomechanical and neurophysiological data collected included ground reaction forces (GRF), kinematics of the trunk and lower limbs, raw EMG from 12 lower-limb muscles, and segmental spinal motor output estimated from spinal mapping.*

Measurement systems and data collection

Participants were asked to walk on a motorized treadmill (H/P Cosmos-Stellar, Germany; belt surface: 1.6 × 0.65 m) instrumented with four force transducers (Arsalis®, Belgium) mounted underneath its frame. These sensors recorded the three components of the ground reaction forces (GRF): in the vertical (F_v), fore-aft (F_f), and mediolateral (F_l) (Willems & Gosseye, 2013) direction. The GRF data were collected at a sampling rate of 1000 Hz

Surface electromyographic activity (EMG) of muscle was recorded from 12 muscles on the right lower-limb using a Delsys Trigno Wireless System (Delsys®, Boston, MA), at a sampling frequency of 2048 Hz. The muscles recorded included *erector spinae* (ES), *tensor fasciae latae* (TFL), *gluteus medius* (Gmed), *vastus medialis* (VM), *vastus lateralis* (VL), *rectus femoris* (RF*), tibialis anterior* (TA), *semitendinosus* (ST), *biceps femoris* (BF), *gastrocnemius lateralis* (GL), *gastrocnemius medialis* (GM), and *soleus* (SOL). Before placing the electrodes, the skin was prepared (shaved and clean) in line with SENIAM recommendations (www.seniam.org). Electrodes were placed parallel to muscle fibers after palpation, and the signal quality was verified during isolated contractions. EMG signals were synchronized with kinetic and kinematic data using a Delsys Trigger Module.

The three-dimensional full-body kinematic data were collected at 240 Hz using a Qualisys motion capture system (Gothenburg, Sweden), which included 14 Oqus 600+ and 4 Miqus M3 cameras positioned around the treadmill. A total of 20 reflective markers were attached bilaterally to anatomical landmarks including the neck, shoulders, elbows, wrists, anterior and posterior superior iliac spines, greater trochanters, external condyles of the knee, malleoli, and fifth metatarsal heads, following the configuration of the Qualisys sports marker set.

Data analysis

General gait parameters: The first instant of foot contact (FC) and toe-off (TO) events were estimated from the displacement of the center of pressure (CoP) on the treadmill belt (Meurisse et al., 2016a). FC and TO were automatically identified based on the computation of *dpath*, which represents the difference between the CoP trajectory and a straight-line approximation connecting its initial and final positions within a step cycle (Meurisse et al., 2016a). A stride was delimited by two successive right FC. Contact phases were measured as the time between foot contact (FC) and toe-off (TO) of the same leg. Swing time was calculated as the duration from TO to the subsequent FC of the same leg.

Kinetics: The fore-aft and vertical velocity of the CoM were calculated from the respective components of the GRF, following the procedure described by Dewolf et al.(Dewolf et al., 2016). In short, the fore-aft acceleration of the CoM was calculated as $a_f = F_f/m$, where m is the subject's body mass. The vertical acceleration of the CoM was calculated as $a_v = (F_v - m\, g)/m$, where g is the acceleration of gravity. The vertical (V_v) and the forward velocity (V_f) of the CoM were calculated by time-integration of a_v and a_f, respectively, plus an integration constant, which was computed so that the average velocity over a stride was equal to zero. The vertical and forward displacements of the CoM (S_v and S_f, respectively) were then computed by time-integration of V_v and V_f. The time of occurrence of the minimum vertical velocity of the CoM ($V_{v,min}$) at the beginning of the double contact phase was detected. Additionally, the peak vertical forces under the front (F_{front}) and back (F_{back}) legs were identified during DC

Kinematics: Kinematic recordings were resampled to 1000 Hz to align temporally with kinetic data. Segment angles (including the trunk, thigh, shank, and foot) were calculated relative to the vertical (Borghese et al., 1996; A. H. Dewolf et al., 2018), where 0° indicated vertical alignment. Hip, knee, ankle and trunk joint angles were then computed from the relative orientations of adjacent segments. Each gait cycle was time-normalized by interpolating it to 400 points. For each participant, segment angles were averaged across all strides. From these, we extracted the mean angle and range of motion

(ROM) for the hip, knee, and ankle joints, as well as trunk elevation, both the range of motion (ROM) and mean values were calculated across the gait cycle.

EMG: The 12 raw EMG signals were resampled at 1000 Hz to match the sampling frequency of the kinetic data. The signals were high-pass filtered at 30 Hz, then rectified, and subsequently, low-pass filtered using a zero-lag 4th-order Butterworth filter at 10 Hz. Gait cycles containing evident artefacts, such as signal loss or saturation due to sensor detachment, were manually excluded following visual inspection. The time scale was normalized by interpolating each gait cycle to 400 points. The root mean square (RMS) of each EMG signal was then computed as a measure of overall muscle activation amplitude across the gait cycle. For each condition, the FWHM was determined as the duration during which the EMG activity exceeded half of its maximum value (Martino et al., 2014; Santuz et al., 2020). The centre of activity (CoA) of each EMG signal was measured as an estimation of the timing of activation. The CoA during the gait cycle was calculated using circular statistics (Batschelet, 1981) and plotted in polar coordinates (polar direction denoted the phase of the gait cycle, with angle α that varies from 0 to 360°).The CoA of the EMG waveform was calculated as the angle of the resultant vector (i.e., the first trigonometric moment) pointing to the center of mass of the circular distribution of EMG activity relative to the gait cycle (Martino et al., 2014), computed as:

$$A = \sum_{i=1}^{400} (\cos \alpha_i X \ EMG_i),$$

$$B = \sum_{i=1}^{400} (\sin \alpha_i X \ EMG_i),$$

$$CoA = tan^{-1}(B/A)$$

where EMG_i represents the EMG amplitude at each phase of the gait cycle. The CoA was preferred, as identifying a clear peak of activity was impractical for most muscles (Martino et al., 2014).

The EMG activities were then mapped onto the rostro-caudal positions of the motoneuron (MN) pools in the human spinal cord, ranging from segments L2 to S2. This mapping followed the myotomal charts of Kendall et al. (F.

Kendall et al., 2005), based on the approximate rostro-caudal location of MN pools innervating different muscles in the human spinal cord. In general, each muscle is innervated by several spinal segments. To reconstruct the output pattern of any given spinal segment S_j, all rectified EMG waveforms corresponding to that segment were averaged as:

$$S_j = \frac{\sum_{i=1}^{n_j} k_{ij} \cdot EMG_i}{n_j}$$

where n_j is the number of EMG_i waveforms corresponding to the j^{th} segment, k_{ij} is the weighting coefficient for i^{th} muscle (from Kendall's chart). The assumption implicit in this methods is that the rectified EMG provides an indirect measure of the net firing of MNs of that muscle in the spinal cord. To compute the total motor output for each condition, the motor output patterns across the gait cycle were summed across the lumbar, sacral and all spinal segments. The mean activation of the lumbar (L2 to L5) and sacral (S1 to S2) segments was computed by averaging the motor output patterns for each region. Finally, the FWHM and the CoA were calculated for both lumbar and sacral segments. The co-activation index (CI) was assessed between the lumbar and sacral segments using the following formula (Mari et al., 2014; Rudolph et al., 2000):

$$\mathrm{CI} = \frac{\sum_{j=1}^{400}\{[\mathrm{lumbar}_j + \mathrm{sacral}_j]/2\} \mathrm{x} [\mathrm{Lumbar}_j/\mathrm{sacral}_j]}{400},$$

where sacral and lumbar represent the mean activation of the segments during the stride. The CI was then averaged over the entire gait cycle (j = 1:400), providing an estimate of the co-activation over the entire cycle. High CI represent a high level of activation of both spinal segments across a large time interval, whereas low CI indicate either a low-level activation of both segments or a high-level activation of one segment along with low-level activation of the other one. Spinal-level co-activation provides a functional representation of neuromuscular output, as previously used to assess age-related co-activation (Dewolf et al., 2021b).

Statistics

For each dependent variable, including spatiotemporal measures, step-to-step transition kinetics, segment elevation and joint kinematic angles, muscle activation metrics, and spinal motor outputs, the normality of residuals was visually assessed using Q–Q plots. When the assumption of normality was not met, a log10 transformation was applied, and normality was re-evaluated and confirmed. A linear mixed-effects model was conducted with age group (young vs. older adults) and tsDCS condition (stimulation *vs*. sham) as fixed factors, and time (pre vs. post) as a repeated measure. Interaction terms (e.g., Age × tsDCS × Time) were included, and post hoc pairwise comparisons were performed using Bonferroni test. Additionally, Statistical Parametric Mapping (SPM) was used to assess the F_v and the spinal motor output of lumbar and sacral segments. In this analysis, statistical significance was defined based on a critical threshold $t^* > 4.3$, which the SPM{t} statistic must exceed at any point in the gait cycle to detect a meaningful difference. All statistical analyses were conducted using GraphPad Prism 8.0.1, with an alpha level of 0.05. SPM analyses were performed in MATLAB R2023a using the open-source spm1d toolbox (v0.4.12).

Results

Some variability in participant characteristics was observed between groups (Table 4.1). As expected, a significant difference in age confirmed the separation between young and older adult groups ($F_{1,34} = 1011$, $p < 0.0001$). Older adults had a lower average height compared to young adults ($F_{1,35} = 12.14$, $p = 0.0013$), with no effect of tsDCS or interaction ($p > 0.16$). No main effect of age or tsDCS was found on body weight (all $p > 0.41$). However, an interaction between age and tsDCS was observed ($F_{1,35} = 5.822$, $p = 0.0212$). BMI was slightly lower in young adults ($F_{1,35} = 4.950$, $p = 0.0326$), with a significant age and tsDCS interaction ($p = 0.0326$). No main effect of tsDCS was observed ($p = 0.5709$). Post hoc comparisons showed a trend toward higher BMI in YA sham compared to YA active ($p = 0.0527$). Finally, no significant differences were found in MET-min/week across age, tsDCS, or their interaction (all $p > 0.43$).

Table 4.1. Participant characteristics grouped by age and stimulation condition (sham or stim).

	Young adults		Older adults		
Variable	sham	stim	sham	stim	
n	10	10	9	10	
Sex	♂=8, ♀=2	♂=9,♀=1	♂=3,♀=6	♂=5,♀=5	
Age (years)	27.2 ± 4.9	27.3 ± 3.3	71.4 ± 4.1	71.6 ± 5	+
Height (cm)	177.7 ± 7.6	179.9 ± 8.3	166 ± 5.5	171.9 ± 12.2	+
Weight (kg)	76.3 ± 7.3	71.4 ± 10.4	66.7 ± 7.2	76.1 ± 14.9	3
BMI (kg m^{-2})	24.2 ± 2.1	22 ± 2.4	24.2 ± 2.7	25.5 ± 2.6	3 +
Met-min/week	3888 ± 2650.4	4968 ± 1614.7	4651.1 ± 6063.7	3926 ± 2852	

***Table 4.1** presents the characteristics of the young and older adult participants, grouped according to stimulation condition (sham or stim). The variables include age (years), height (cm), weight (kg), body mass index (BMI) (kg/m²), and physical activity level expressed as MET-min/week. Data are reported as means with corresponding standard deviations (±). Significant main effects of age and the interaction between age and tsDCS are indicated with the symbols + and ³, respectively.*

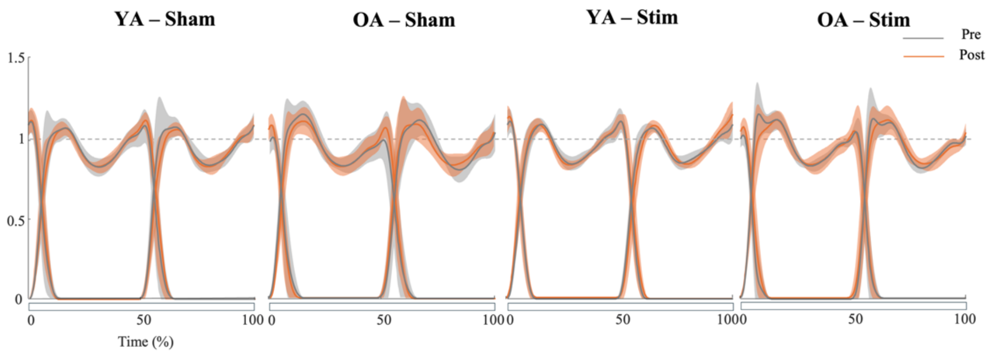
Walking Kinetics

In all participants, the F_v curve over a gait cycle showed the characteristic double-peak so-called M-pattern during walking (Fig. 4.2A). Notably, SPM{t} analysis revealed no significant pre–post differences in any group, with all test statistics remaining below the critical threshold ($t > 4.3$; all $p > 0.05$).

As already documented, in older adults, F_{back} was significantly lower ($F_{1,36} = 9.307$, $p = 0.0043$) whereas F_{front} was significantly greater ($F_{1,36} = 23.96$, $p < 0.0001$), as compared to young adults. No effects of tsDCS or interaction were observed on F_{back} (all $p > 0.23$) but the participant who received the tsDCS stimulation displayed a higher F_{front} ($F_{1,34}$ = 38.82, p < 0.0001) without interaction with age (all p > 0.30). Finally, $V_{v,min}$ occurred later in the gait cycle in older adults as compared to young ($F_{1,36}$ = 29.66, p < 0.0001), and later in the sham group ($F_{1,34}$ = 6.627, p = 0.0146), with no interaction effects (Fig. 4.2C).

Significant differences in spatiotemporal gait parameters were found across groups (Fig. 4.2C). Contact time was significantly shorter in older adults compared to young participants ($F_{1,36}$ = 22.85, p < 0.0001). Similarly, swing time ($F_{1,36}$ = 6.408, p = 0.0159) and stride time ($F_{1,36}$ = 13.37, p < 0.0001) were both reduced in older adults. A significant three-way interaction was found for swing time (age × time × tsDCS; $F_{1,34}$ = 4.339, p = 0.0448). No other significant main or interaction effects were found (all p > 0.22).

A



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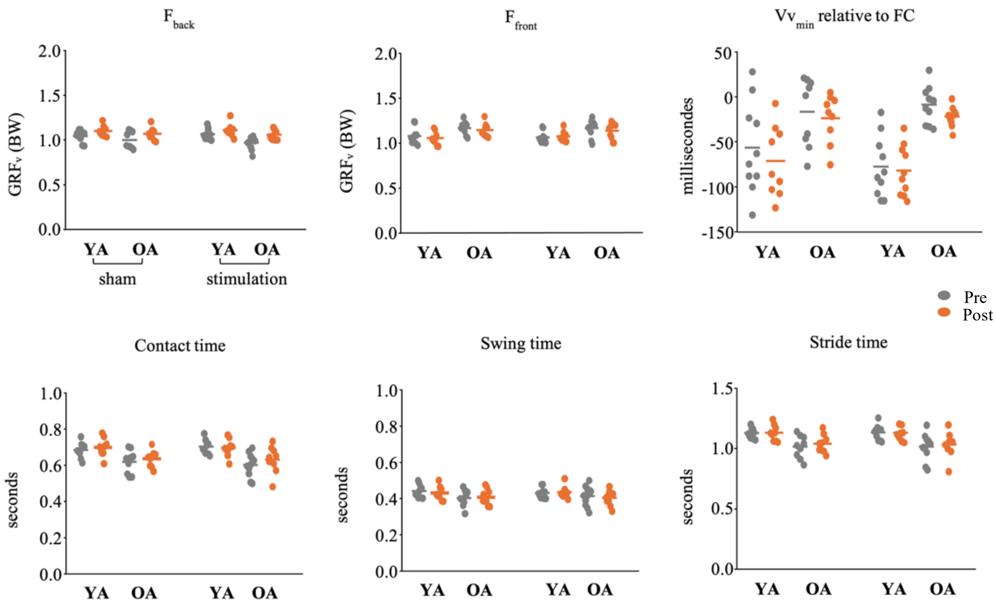


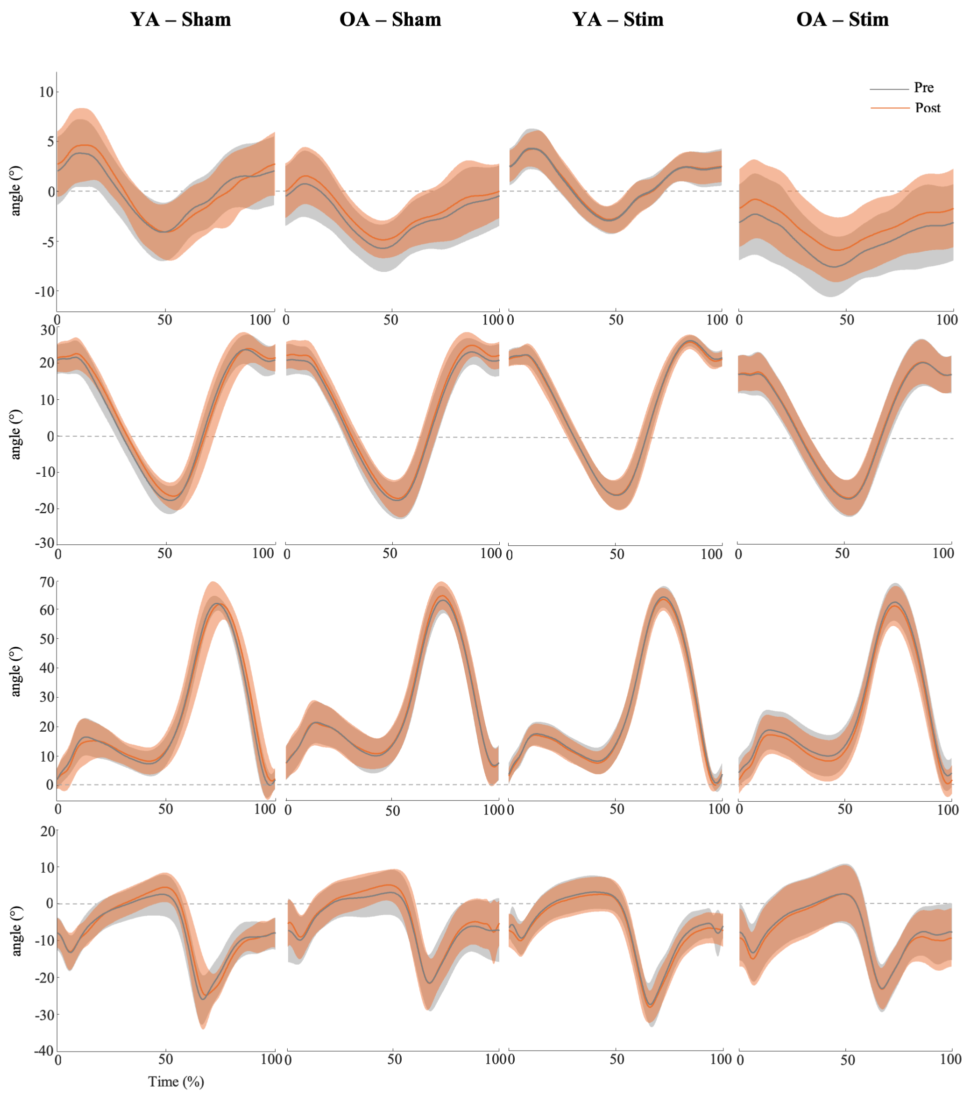
Fig. 4.2. *Vertical ground reaction force (GRF$_v$) patterns and summary metrics before (Pre) and after (Post) trans-spinal direct current stimulation (tsDCS) across age groups. (**A**) Time-normalized vertical GRF signals (mean ± SD) over five gait cycles, expressed relative to body weight, and separated by group and condition. (**B**) Individual and group-averaged results for GRF$_v$ metrics, including first and second peak forces, loading rate, and impulse, grouped by condition (Pre/Post), age (YA/OA), and stimulation (sham/stim). Gray and red denote sham and stimulation, respectively. Asterisks indicate statistically significant main effects or interactions (p < 0.05).*

Kinematics of walking

Segments and joint kinematics showed group- and intervention-related differences primarily in trunk and hip motion (Fig. 4.3A). Specifically, an increased trunk elevation was significantly different between age-group ($F_{1,36}$ = 21.42, p < 0.0001) and between tsDCS condition ($F_{1,35}$ = 4.697, p = 0.0371), with an additional interaction between age and tsDCS ($F_{1,35}$ = 4.508, p = 0.0409). Indeed, older adults reduced their trunk inclination after the tsDCS ($F_{1,35}$ = 4.5, p = 0.04). This effect was confirmed when analyzing the minimum trunk flexion angle (peak flexion), which also showed a significant main effect of Age ($F_{1,35}$ = 11.55, p = 0.0017), tsDCS ($F_{1,35}$ = 6.564, p = 0.0149), and their interaction ($F_{1,35}$ = 5.422, p = 0.0258). Post hoc analysis revealed that minimal trunk flexion significantly decreased in active older adults after stimulation (p = 0.0168) (Fig. 4.4A). No other main effects or interactions reached significance for ankle, knee, or hip angles (all $p > 0.10$), except a main effect of tsDCS on hip angle ($F_{1,35}$ = 4.738, p = 0.0363) (Fig. 4.3B).

To assess time-resolved differences in joint kinematics, SPM{t} was applied across the gait cycle. In young and older adults with both sham and stimulation, no significant differences were found with time in any of the angle analyzed (p > 0.05).

A



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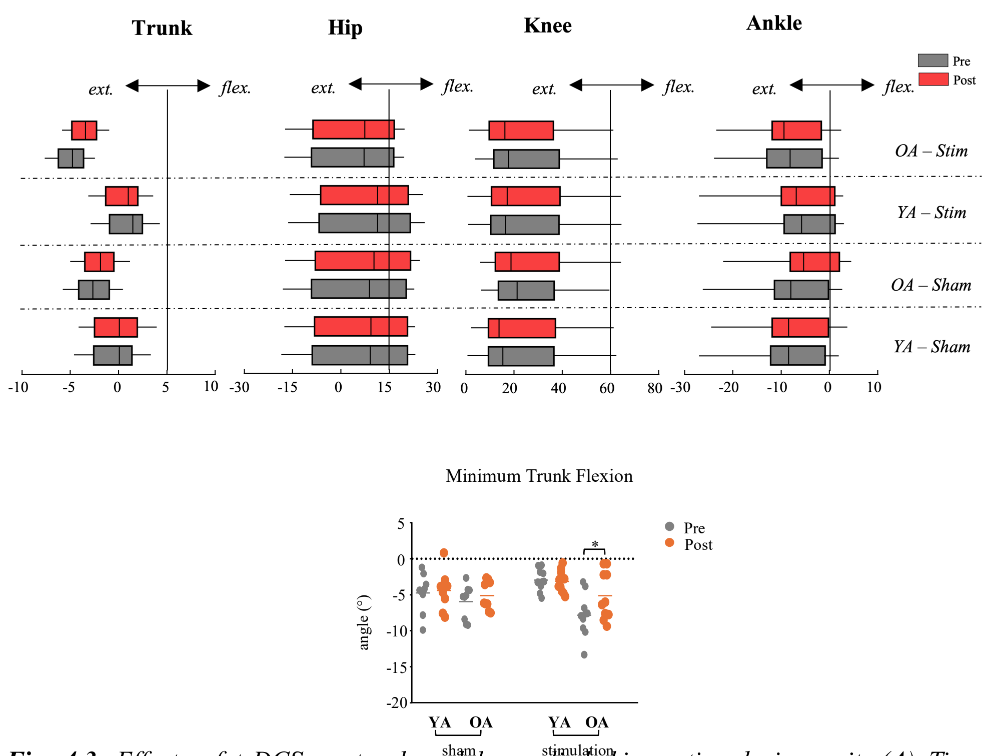


*Fig. 4.3. Effects of tsDCS on trunk and lower limb kinematics during gait. (**A**) Time-normalized angular trajectories (Mean ± SD) of the trunk, hip, knee, and ankle across the gait cycle (%), shown for young adults (YA) and older adults (OA) under sham and stimulation (Stim) conditions, before (gray) and after (red) the intervention. Joint angles are represented in the sagittal plane, with extension (ext.) and flexion (flex.). (**B**) Range of motion of joint angles during the gait cycle for each joint and group/condition. (**C**) Minimum trunk flexion angle (°) pre - and post-intervention. The asterisk (*) denotes statistically significant differences at p < 0.05.*

Muscle activation during walking

The EMG activation amplitudes across the gait cycle revealed group-specific differences in muscle activation amplitude (Fig. 4.4). Differences related to age were found for VL ($F_{1,35} = 6.441$, $p = 0.0158$), RF ($F_{1,34} = 29.40$, $p < 0.0001$), TA ($F_{1,35} = 12.39$, $p = 0.0012$), ST ($F_{1,35} = 7.567$, $p = 0.0093$), BF ($F_{1,34} = 33.40$, $p < 0.0001$), and GM ($F_{1,35} = 12.39$, $p = 0.0012$). Effects of tsDCS were observed for VM ($F_{1,30} = 6.351$, $p = 0.0173$), RF ($F_{1,34} = 6.089$, $p = 0.019$), TA ($F_{1,35} = 10.09$, $p = 0.0031$), ST ($F_{1,35} = 4.472$, $p = 0.0417$), BF

($F_{1,35} = 9.905$, $p = 0.0035$), GL ($F_{1,35} = 22.16$, $p < 0.0001$), GM ($F_{1,35} = 10.09$, $p = 0.0031$), and SOL ($F_{1,33} = 9.591$, $p = 0.004$). No significant differences were found for ES, TFL, or GMED (all $p > 0.05$) (Fig. 4.5). It is important to note that these differences reflect only changes in activation amplitude (RMS) (Fig. S4.1).

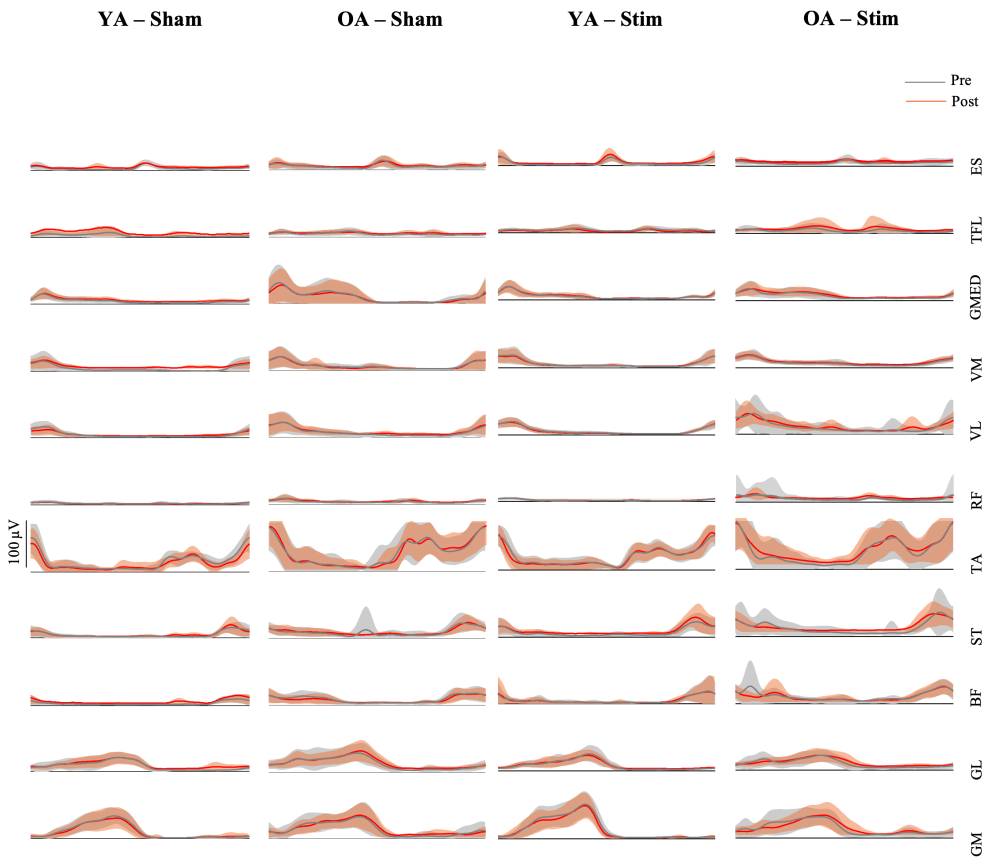
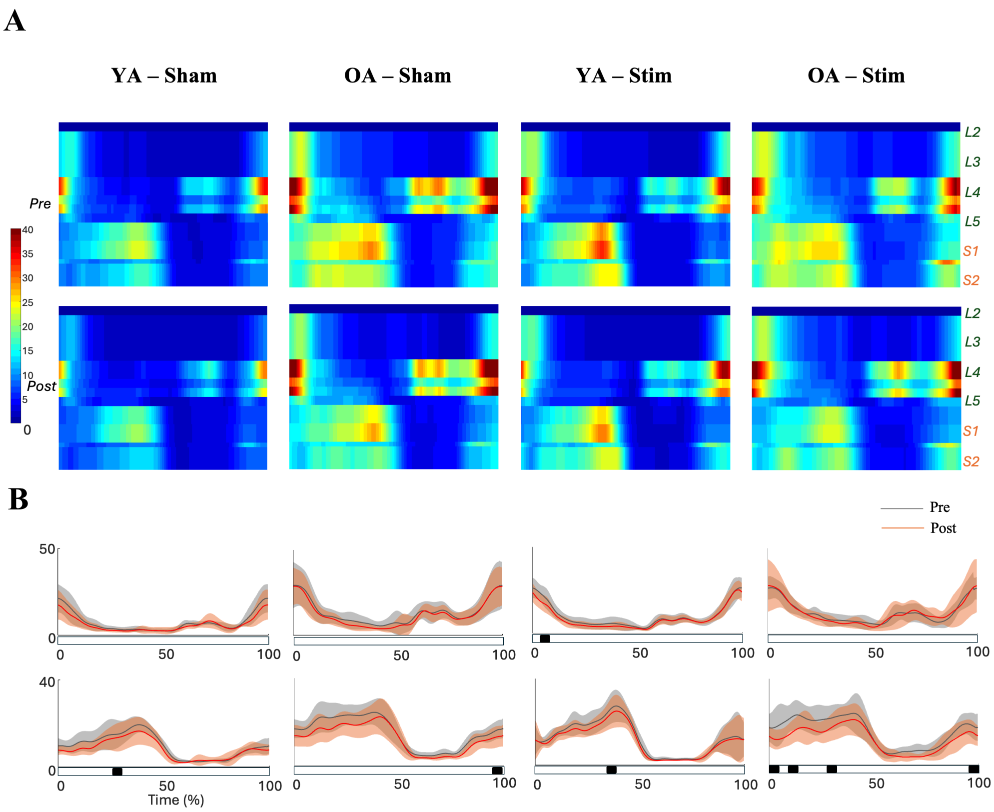


Fig. 4.4. *Rectified EMG profiles of 12 lower limb muscles across the gait cycle for each experimental condition. Each subplot represents the average signal (mean ± SD) for a given muscle and condition, expressed over 100% of the gait cycle. Conditions are arranged by group and stimulation: YA-Sham, OA-Sham, YA-Stim, and OA-Stim (left to right). EMG activity is shown for both pre-intervention (black) and post-intervention (red) sessions. Shaded areas represent inter-subject variability (±1 SD). EMG amplitude is shown in arbitrary units, normalized to each subject's peak value.*

When the EMG are mapped onto the rostro-caudal positions of the MN pools, the spinal maps revealed changes in the spinal distribution of motor activity across the gait cycle with age and stimulation (Fig. 4.5A). The lumbar and sacral motor outputs were compared across the gait cycle to assess the level of activation before and after the stimulation (or sham). The SPM{t} analysis revealed a significant increase in lumbar output in young adults with tsDCS between 0–5% of the gait cycle (p = 0.006, t = 4.49). No other significant changes were found in lumbar output for the remaining groups (all $p > 0.05$). In contrast, sacral activity showed several significant differences. The young adults after sham exhibited a difference between 25–30% of gait cycle (p = 0.039. t = 4.71). The young adults after a tsDCS showed a difference between 36–44% (p = 0.011, t = 4.65) of gait cycle. The older adults with sham showed a significant difference at 98–100% (p = 0.035, t = 4.24). More importantly, the older adults after tsDCS revealed three significant intervals: 95–3%, 7–9%, and 32–36% (all $p < 0.05$, t = 4.37).



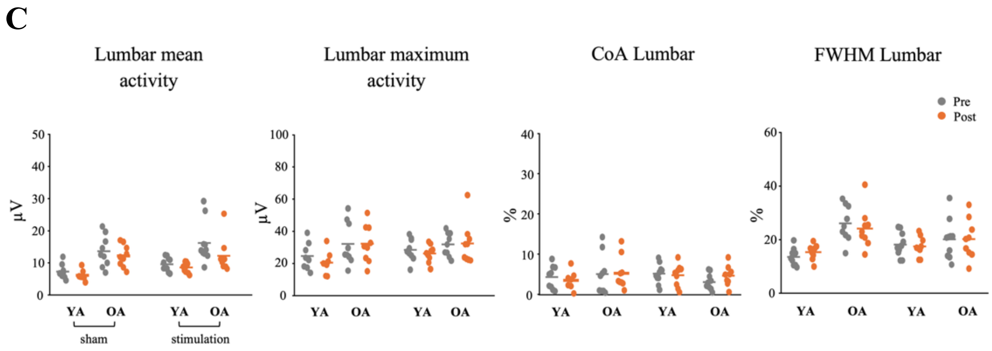


Fig. 4.5*. Spinal motor output and summary parameters before and after tsDCS. (**A**) Interpolated color maps of spinal motor output across lumbar (L2–L5) and sacral (S1–S2) segments over the normalized gait cycle (0–100%) for each group and condition (YA-Sham, OA-Sham, YA-Stim, OA-Stim). Color intensity reflects relative segmental activation levels. (**B**) Ensemble-averaged time-series (mean ± SD) of lumbar and sacral segmental output, shown separately for pre (black) and post (red) stimulation conditions. Black rectangles in the lumbar and sacral activation time-series indicate regions of the gait cycle where significant differences between pre and post conditions were identified by statistical parametric mapping (SPM). (**C**) Summary parameters of spinal activation per participant, including maximum amplitude, mean amplitude, center of activity (CoA), and full width at half maximum (FWHM), expressed as a percentage of the gait cycle. Gray and red dots represent pre and post values, respectively; horizontal bars indicate the group mean respectively; horizontal bars indicate group means. No statistically significant differences were observed.*

To further characterize spinal motor output across the gait cycle, mean and maximum activity were analyzed at the lumbar and sacral levels (Fig. 4.6B). For mean activity, a main effect of age was observed at both the lumbar ($F_{1,34}$ = 21.57, $p < 0.0001$) and sacral levels ($F_{1,34}$ = 18.94, p = 0.0001), with older adults showing higher average amplitudes. A significant effect of tsDCS was also found at both lumbar ($F_{1,31}$ = 17.17, p = 0.0002) and sacral segments ($F_{1,31}$ = 14.51, p = 0.0006), and a significant main effect of time was present only in the sacral region ($F_{1,34}$ = 5.430, p = 0.0259), indicating an increase in mean output following stimulation. Maximum activity showed a significant main effect of age at the lumbar level ($F_{1,31}$ = 5.395, p = 0.0269), but not at the sacral level ($F_{1,31}$ = 2.153, p = 0.1524). tsDCS significantly increased sacral maximum activity ($F_{1,31}$ = 9.716, p = 0.0039), and a significant age × tsDCS interaction was observed ($F_{1,31}$ = 4.467, p = 0.0427) (Fig 5.5C), suggesting a differential responsiveness in older adults at the sacral level.

Temporal organization of spinal output was assessed through the timing and duration of segmental activity. The CoA of sacral activation occurred earlier in older adults compared to young adults ($F_{1,34} = 9.890$, $p = 0.0034$), with no difference at the lumbar level ($p > 0.05$). Burst duration, assessed by FWHM, was greater in older adults at both lumbar ($F_{1,34} = 13.15$, $p = 0.0009$) and sacral segments ($F_{1,34} = 4.239$, $p = 0.0472$), with an age–time interaction observed at the lumbar level ($F_{1,34} = 6.040$, $p = 0.0192$) (Fig 5.5C).

Discussion

This study evaluated whether a single session of anodal tsDCS over the lumbosacral region could modulate gait in young and older adults. We analyzed spatiotemporal parameters, segment and joint kinematics, and the spinal motor output estimated from multi-muscle recording. Our findings partially supported our hypothesis: tsDCS modulated sacral output in older adults, with significant changes across the gait cycle and subtle kinematic adaptations, particularly in trunk motion.

It is well documented that aging is associated with specific neuromuscular adaptations during gait (Dewolf et al., 2021a, 2021b; Ivanenko et al., 2013; Monaco et al., 2010; Nùñez-Lisboa et al., 2023). Our findings reinforce previous observations showing that older adults exhibit an altered spinal motor output, characterized by a wider and earlier activation of the sacral motor pools compared to younger adults (Dewolf et al., 2021a, 2021b). This different of sacral activation timing observed (Fig. 4.5A) likely reflects an earlier recruitment of distal muscles, such as plantar flexors, which has been consistently reported as a hallmark of the aging gait pattern (Boyer et al., 2017; Franz, 2016; Franz & Kram, 2013). This premature activation is thought to contribute to the altered force profile observed in older adults, with reduced propulsion (lower F_{back}) and greater anterior collision (higher F_{front}) during gait (Fig. 4.1B), consistent with previous results (Dewolf, Meurisse, et al., 2019; Dewolf et al., 2022). The earlier activation of sacral segments induces a reduced delay between lumbar and sacral motor pool bursts(Monaco et al., 2010; Nùñez-Lisboa et al., 2023), likely reflecting a compensatory mechanism to maintain stability in the context of degraded afferent input.

Following tsDCS, older adults exhibited a reduced maximal and average trunk inclination (Fig. 4.3C). Anterior trunk flexion is a well-recognized hallmark of aging gait and has been interpreted as a compensatory response to diminished propulsive capacity and postural control (Dewolf et al., 2022; Honda et al., 2023). This finding may have significant clinical implications, since in a previous work (Núñez-Lisboa et al., 2024), we showed that greater trunk inclination is associated with reduced propulsive force (F_{back}) and higher collision force (F_{front}), and change in gait kinematics and spinal motor output. Notably, the increased trunk flexion has also been linked to higher mechanical and metabolic cost during walking (Delabastita et al., 2021; Dewolf et al., 2022), where postural alignment plays a critical role in step-to-step weight transfer (Núñez-Lisboa et al., 2024). Therefore, the subtle but significant effect of re suggests that spinal neuromodulation may contribute to improving postural and in turn gait pattern in older adults.

Based on the relation between trunk inclination (Núñez-Lisboa et al., 2024), we expected that the change in trunk inclination would have induced changes in gait kinematics, kinetics and spinal motor output. While no significant changes were observed on kinetics (Fig. 4.2) and joint kinematics (Fig. 4.3) following tsDCS, older adults exhibited changes in the sacral burst of activation. Indeed, after tsDCS, a reduction of the amplitude of sacral motor output is observed, particularly during early and late stance phases (Fig. 4.5A), but without changes in temporal organization (Fig. 4.5C). While a modest decrease in sacral output is also observed in the sham group near the end of the gait cycle, the reduction following tsDCS after stimulation is more robust, spanning both early and late stance (Fig. 4.5A). Given that sacral motor pools innervate distal muscles that contribute to forward propulsion (F. Kendall et al., 2005), the spinal-level modulation observed in the sacral maps may represent an underlying neural adjustment contributing to this mechanical change. However, the single session of stimulation seems to be not sufficient to induce clear and significant changes of gait biomechanics. Nevertheless, the present results highlight the responsiveness of spinal motor circuits to non-invasive neuromodulation in older adults.

The mechanisms through which tsDCS exerts its effects on gait control are not yet fully understood, but accumulating evidence suggests that spinal neuromodulation can influence both segmental and intrinsic spinal pathways involved in locomotion. The tsDCS is thought to modulate spinal excitability by targeting calcium conductance and persistent inward currents (PICs) that augment motoneuron firing in the spinal cord (Baczyk, M., 2019; Jankowska, 2017). Anodal stimulation has been shown to depolarize afferent terminals and enhance excitatory transmission within spinal circuits, thereby facilitating sensorimotor integration (W. Song & Martin, 2017). This has been supported by evidence showing that tsDCS prolongs the post-exercise depression of the H-reflex, indicating a potentiation of spinal inhibitory mechanisms (Awosika et al., 2019). It is nevertheless important to emphasize that while tsDCS is sometimes referred to as ‘spinal stimulation’, its effects are not attributable to an anodal blockade of axonal conduction in the spinal cord, which involves the inhibition of signal transmission along axons due to sustained anodal current. Instead, evidence suggests that tsDCS modulates the processing of sensory input within spinal circuits, thereby influencing spinal network activity indirectly (Lenoir et al., 2018). This aligns with recent perspectives that tsDCS does not directly depolarize spinal motoneurons but rather influences their excitability via altered sensory inflow (Lenoir et al., 2018; Tajali et al., 2024). These neuromodulatory effects are thought to interact with spinal CPGs, which orchestrate the rhythmic and phase-specific activation of motor pools during gait (Kiehn, 2006). In this context, the observed modulation of sacral output following tsDCS may reflect an acute reorganization of spinal inter-neuronal circuits, particularly those controlling distal musculature involved in propulsion.

Limitations

A key limitation of the present study is the absence of direct measures of spinal excitability, such as H-reflex or TMS assessments, which would have provided more conclusive evidence of spinal involvement (Awosika et al., 2019; Yamaguchi et al., 2020). These findings should be interpreted with caution, as the effects of a single session may not translate into lasting change in spinal cord excitability and functional benefits. Future studies should evaluate repeated tsDCS sessions combined with task-specific gait training to determine whether cumulative adaptations occur. Another limitation of our study is the absence of direct measurements of subcutaneous tissue thickness at the stimulation site, and we assumed comparable fat distribution and thus similar depths of neural structures beneath the skin. However, differences in local adiposity could affect the proportion of current reaching deeper tissues, introducing variability in the neuromodulatory effects of tsDCS (Fernandes et al., 2018). Also, the relatively low stimulation intensity (2.5 mA), which, although within established safety thresholds, may have been insufficient to induce more pronounced or widespread neuromodulatory effects (Eberhardt et al., 2023).

While kinematic changes remained modest and no significant kinetic adaptations were observed, the neuromodulatory effects at the spinal level - particularly in sacral motor pool activity - support the hypothesis that spinal circuits remain plastic in aging. These findings open promising avenues for future interventions aiming to restore more efficient gait patterns and reduce fall risk in older individuals. Future studies should investigate the cumulative and long-term effects of repeated tsDCS sessions, possibly in combination with gait training or task-specific rehabilitation programs, to assess whether spinal neuromodulation can promote durable functional gains.

Conclusion

This study provides preliminary evidence that anodal tsDCS applied over the sacral region can modulate spinal motor output and induce neuromechanical adaptations during gait in older adults. Although the observed effects were moderate, they included relevant changes in postural control and propulsive force generation, aligning with known age-related gait deficits. These findings highlight the potential of tsDCS as a non-invasive tool to address spinal-level motor impairments in aging. Future research should investigate the effects of repeated sessions, long-term adaptations, and functional outcomes to fully assess the therapeutic value of this approach in promoting gait efficiency and stability in older populations.

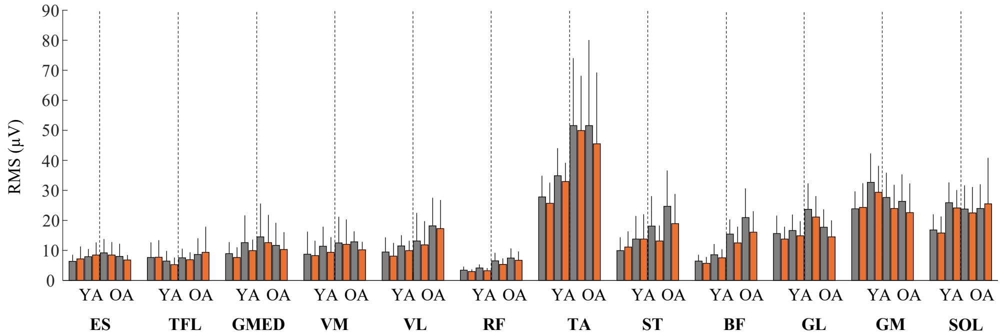


Fig. S4.1. *Individual root mean square (RMS) values across muscles and conditions. RMS amplitudes (µV) are shown for each recorded muscle before (gray bars) and after (red bars) the tsDCS intervention, separated by experimental group. Each bar represents the mean RMS value for one participant and one condition. Muscles are organized along the x-axis according to their anatomical grouping.*

Acknowledgements

The authors express their gratitude to Antoine Samain and Nemo Lecoubet for their help during the experiments

Chapter V
General Discussion

Human locomotion is a complex, adaptive process that emerges from the continuous interaction between neural commands, muscular output, and mechanical constraints. With advancing age, this dynamic system undergoes multifactorial changes that affect mobility, stability, and functional independence. While the mechanical consequences of aging on gait and balance have been well documented, the underlying neuromechanical mechanisms remain incompletely understood. This thesis aimed to bridge this gap by investigating how altered posture (Chapter II), physical activity (Chapter III), and targeted spinal neuromodulation (Chapter IV) shape the neuromechanical control of gait and balance in older adults. Through a series of experiments combining (among others) kinematic, kinetic, and EMG data, this work provides new insights into both the deterioration and plasticity of locomotor function across the adult lifespan.

The effect of age on gait is well-documented in the literature. In all the chapters of this thesis, aging was consistently associated with altered segmental coordination, increased mechanical cost, reduced energy recovery, and less efficient transitions during the double contact phase (Boyer et al., 2017, 2023; Delabastita et al., 2021; A. H. Dewolf et al., 2018; Meurisse et al., 2019b, 2019a; Winter, Patia, et al., 1990). These impairments were reflected in altered spatiotemporal organization (e.g., shorter steps and higher frequency), elevated W_{ext}, delayed CoM redirection, reduced joint mobility, and a more anteriorly inclined trunk posture, which may shift the alignment of the CoM and compromise transition dynamics (Chapter II & III). At the muscular level, older adults demonstrated significantly lower plantar flexor torque than younger adults, a widely reported feature of aging (McPhee et al., 2016; Power et al., 2010) and confirmed in our results (Chapter IV). However, these strength deficits also coincided with broader impairments in neuromuscular coordination, including altered timing and magnitude of muscle activation. This prompted further exploration of spinal-level organization as a determinant of locomotor control. Critically, older adults exhibited age-related changes in spinal output, including a broader temporal distribution of activation and reduced rostro-caudal differentiation, particularly at the sacral level. These alterations, detailed in Chapter III & IV, suggest a decline in segmental motor selectivity and timing precision, consistent with previous reports of age-related reorganization in spinal circuits (Dewolf et al., 2021b; Monaco et al., 2010).

Collectively, these findings confirm the overarching aim of the thesis: to investigate how aging and its modulation through physical activity and spinal neuromodulation shape the neuromechanical control of gait and balance. The results offer new insight into the age-related deterioration of locomotor strategies, while also highlighting the potential for plasticity within the neuromuscular and postural systems. Together, these findings contribute to a more integrated understanding of the mechanisms underlying gait control in older adults.

The Functional Impact of Trunk Inclination

To deepen the integrated understanding of gait control in older adults, we examined how anterior trunk inclination, commonly observed in this population (Honda et al., 2023), affects locomotor efficiency. This postural deviation appears specific to walking and is not typically present in quiet standing, suggesting that trunk inclination may be a task-dependent adaptation rather than a constant postural deficit. From a mechanical standpoint, walking operates through an inverted pendulum mechanism that enables partial energy transfer between E_k and E_p, optimizing muscular efficiency through timely redirection of the CoM (Cavagna et al., 1977; Willems et al., 1995). However, excessive forward displacement of the upper body may misalign the CoM relative to the base of support, potentially increasing the mechanical cost of walking (Aminiaghdam et al., 2016; Grasso et al., 2000). This was confirmed in Chapter II, where experimental trunk flexion in healthy young adults led to reduced energy recovery, elevated W_{ext}, and a greater prevalence of non-anticipated transitions. These mechanical alterations were accompanied by changes in segmental kinematics and prolonged activation of extensor muscles. Although ES activity was not evaluated during the trunk flexion experiment, other chapters in this thesis provide converging evidence that older adults walk with greater trunk inclination and exhibit elevated ES activation. This pattern likely reflects a compensatory neuromuscular response to maintain postural alignment and forward propulsion under altered mechanical conditions. Notably, these adaptations were consistently observed in the chapters examining the influence of physical activity and spinal neuromodulation. While tsDCS demonstrated some potential to modify trunk posture, its impact on ES activation and broader gait dynamics remained limited. These findings suggest that trunk inclination and axial extensor engagement represent critical components of neuromechanical adaptation in aging, warranting further investigation into whether interventions such as tsDCS can effectively modulate these postural control mechanisms.

Although the anterior trunk inclination observed in our older adult cohort (~3°) was relatively modest, it may still exert biomechanical consequences when interacting with other age-related deficits. Previous findings indicate that older adults exhibiting greater trunk flexion (~9.5°) demonstrate delayed CoM redirection and a more pronounced biomechanical alteration during walking (Dewolf et al., 2022), suggesting that postural alignment beyond a certain threshold can compromise gait efficiency. In the present context, the subtle trunk inclination likely does not exceed this threshold but may nonetheless contribute to altered transition dynamics through its combined effect with reduced muscular strength, sensory degradation, and impaired spinal motor output. These interactions highlight the importance of considering trunk posture not in isolation, but as part of a broader set of neuromechanical adaptations that characterize locomotor control in aging. This interpretation provides a conceptual bridge to the subsequent section, which examines trunk inclination as an active modulator of gait efficiency.

Neurodegenerative Foundations

This multifactorial interaction likely reflects the neurodegenerative substrate of aging, in which motor impairments are not solely attributable to peripheral sarcopenia, but also to progressive deterioration of central nervous system structures. Age-related reductions in grey matter volume (Good et al., 2001), loss of motor cortical and spinal neurons (Doherty, 2003; Henderson et al., 1980), decreased synaptic density (Haug & Eggers, 1991), and compromised white matter integrity (Davis et al., 2009) all contribute to diminished descending motor drive (Yue et al., 1999). These central alterations, together with peripheral neuromuscular degradation, can impair coordination, reduce motor selectivity, and disrupt timing precision during locomotion (Dewolf et al., 2021a, 2021b, 2019a; Monaco et al., 2010). These findings reinforce the idea that gait control in older adults results from a convergence of multisystemic decline, where posture, mechanical cost, and motor timing reflect deeper alterations in spinal and supraspinal integrity.

In addition to peripheral and segmental spinal factors, age related deterioration of supraspinal command pathways, particularly those essential for locomotion (reticulospinal, vestibulospinal and corticoreticulospinal) deserves explicit consideration. The reticulospinal and vestibulospinal tracts, together with corticoreticulospinal inputs from premotor, supplementary motor and primary motor cortices, constitute the core supraspinal command that couples postural control with step generation (Takakusaki, 2017). With aging, vestibulospinal drive weakens as vestibular end organs degenerate, reducing stabilization of the head and trunk (Iwasaki & Yamasoba, 2015). The reticulospinal contribution at the lumbosacral level is reduced because the spinal target is smaller due to late life loss of anterior horn cells and because the soleus H reflex shows blunted phase dependent modulation during walking, indicating weaker descending control of sensory motor gain (Henderson et al., 1980; Raffalt et al., 2015). Because the corticoreticulospinal command depends on basal ganglia dopamine, cerebellar calibration and brainstem relays (Takakusaki, 2017), age related decline in these systems likely degrades that command, as shown by smaller or delayed anticipatory postural adjustments during gait initiation (Bernard &

Seidler, 2014; Karrer et al., 2017; Ransmayr et al., 2000). Together these changes reduce the effectiveness of the pathway architecture that links postural control to step generation.

Interestingly, these neuromechanical impairments echo patterns observed at the opposite end of the lifespan. As shown in the supplementary chapter of this thesis (*Effect of age and speed on the step-to-step transition strategies in children*) and discussed in Supplementary Chapter I, both toddlers and older adults exhibit a broader and temporally prolonged activation of spinal segments, particularly at the sacral level, alongside a high prevalence of non-anticipated transition strategies during walking. These shared features suggest that immature and degenerated neuromuscular systems face similar challenges in coordinating push-off and redirection of the CoM. From a developmental perspective, locomotor control emerges through a proximal-to-distal progression, with distal motor control maturing later as spinal networks and corticospinal pathways become more refined (Dominici et al., 2011). In contrast, aging appears to follow an inverse trajectory, where the earliest signs of neuromuscular deterioration are often observed in distal segments of the lower limb. This distal vulnerability has been linked to structural degeneration of spinal pathways, reduced corticospinal modulation, and impaired motor unit remodeling in distal musculature (Baudry et al., 2015; Doherty, 2003; Martino et al., 2018).

Together, these lifespan comparisons highlight a fundamental principle: that reduced corticospinal modulation, whether due to immaturity or degeneration, compromises segmental specificity and the timely coordination of neuromuscular transitions during walking. This convergence reinforces the role of spinal output organization as a biomarker of gait integrity, with important implications for early detection and intervention strategies across the lifespan.

Neuromechanical Benefits of Physical Activity in Aging Gait

Physical activity plays a central role in preserving functional mobility with aging, but its influence extends far beyond muscle strength alone. While sarcopenia refers to the loss of muscle mass, and dynapenia to the loss of muscle strength without a corresponding reduction in mass, these have long been considered the primary contributors to mobility decline (Clark & Manini, 2008). Emerging evidence emphasizes that coordinated neuromechanical control is equally shaped by habitual movement behaviors. Importantly, physical inactivity (defined as insufficient engagement in moderate-to-vigorous activity) is conceptually distinct from sedentary behavior, which refers to prolonged periods of low-energy behavior such as sitting or lying down (Tremblay et al., 2017). Both are prevalent in aging populations and independently associated with poor functional outcomes. To mitigate these risks, the World Health Organization recommends that adults engage in at least 150 to 300 minutes of moderate-intensity aerobic activity per week (Bull et al., 2020). However, many older individuals fall short of these thresholds.

Participants classified as more active in this study showed markedly higher engagement in aerobic exercise, with over 93% reporting regular participation compared to just 53% in the less active group (Table 5.1). This contrast highlights that more active individuals not only met, but often exceeded basic physical activity recommendations. Conversely, the less active group was characterized by sporadic or minimal engagement in structured exercise, relying more on general daily activities such as walking. These distinctions provide critical context for interpreting the neuromechanical differences observed in aging gait.

Table 5.1. Distribution of physical activity types among less active and more active participant groups

Group	Number of participants	Strength training (n)	Aerobic training (n)	Sports (n)	General physical activity (n)
OA less active	15	0	8	3	8
OA more active	15	3	14	4	6
YA less active	16	2	1	7	1
YA more active	15	6	10	8	2

Table 5.1 summarizes the distribution of reported physical activity types among participants classified as less or more active in each age group, who reported engaging in different types of physical activity: strength training, aerobic exercise, sport participation, and general physical activity (e.g., yoga, dancing, walking). Participants could report more than one type of activity; thus, column totals may exceed the number of individuals per group.

In this thesis, physical activity level was assessed using the GPAQ, which allowed for the classification of participants into more active and less active groups based on their MET-minutes per week. Although self-report questionnaires are commonly used in population-based studies, they are subject to limitations such as recall bias and subjective estimation. In contrast, objective tools like inertial measurement units, accelerometers, and smartwatches are increasingly used to monitor physical activity with greater temporal and intensity resolution (Troiano et al., 2014). While such tools were not employed in the present work, their use could enhance future assessments of habitual movement behaviors.

Taken together, these findings suggest that physical activity supports a more integrated neuromechanical profile during gait, where temporal coordination, energy efficiency, and postural control coalesce to preserve mobility. Rather than acting on isolated systems, movement habits appear to sustain the interplay between neural, muscular, and mechanical components essential for stable and efficient locomotion in older age.

Spinal Plasticity and Acute Modulation through tsDCS

The results of this thesis provide preliminary evidence that spinal locomotor circuits in older adults retain a degree of plasticity that can be modulated through non-invasive neuromodulation. In Chapter IV, a single session of tsDCS produced subtle improvements in trunk posture during gait, suggesting that spinal excitability and segmental coordination can be transiently influenced by external electrical modulation. However, no measurable changes were observed in transition strategies, muscle activation patterns, or axial extensor engagement. This aligns with previous findings indicating that the behavioral effects of tsDCS may be limited following a single session, and that repeated stimulation may be necessary to enhance motor consolidation and retention (Awosika et al., 2019). These limited acute effects likely reflect subthreshold, state dependent modulation and current distribution between spinal cord, roots, and skin, requiring repeated, task specific sessions.

Despite persistent unknowns about the precise physiological mechanisms within spinal networks, tsDCS has seen growing use in neurorehabilitation and as an adjunct to physical training (Angius et al., 2017; Berry et al., 2017). Converging animal and human evidence indicates that a constant current applied across the spinal axis can transiently reweight synaptic efficacy within descending pathways (reticulospinal, vestibulospinal and corticoreticulospinal), ascending proprioceptive and cutaneous tracts, and local inter-neuronal circuits including CPG and premotor networks. At the final common pathway, anodal tsDCS tends to depolarize spinal motoneurons, lowering recruitment thresholds and increasing spike probability and rate coding, whereas cathodal stimulation tends to hyperpolarize and reduce rate coding gain (Baczyk, M., 2019). At the segmental level, a reproducible signature of anodal tsDCS effects is a reduction of the H reflex, indexing increased presynaptic inhibition of Ia afferents and altered alpha motoneuron pool excitability (Bocci et al., 2015). Accordingly, in Chapter IV we applied anodal tsDCS following a rostro-caudal stimulation order targeting lumbar and sacral segments (Monaco et al., 2010), with the aim of transiently depolarizing spinal motoneurons and facilitating the translation of descending and sensory inputs into motor pool output. Similar improvements in locomotion have been reported with anodal tsDCS combined with backward treadmill training (Awosika et al., 2019).

Several practical factors could have attenuated the effect size described in Chapter IV. Interindividual differences in subcutaneous adipose tissue and vertebral canal geometry can alter current paths and the field that reaches spinal gray matter, so equal stimulator currents do not imply equal in cord electric fields across participants (Fernandes et al., 2018). The 2.5 mA dose, while within safety limits, may also sit near the lower bound for producing robust physiological effects with the present electrode geometry (Eberhardt et al., 2023). In addition, a nontrivial fraction of the dose likely spreads through skin and dorsal roots, modulating cutaneous and root afferents more than intraspinal circuits, which may yield postural adjustments but not immediate changes in gait mechanics. Future studies should individualize dosing to anatomy, include an in vivo segmental physiological measure such as the H reflex to verify engagement during the task, and pair stimulation with task specific gait practice across multiple sessions.

Taken together, our findings support tsDCS as a plausible means to modulate spinal circuits in older adults, while underscoring the need to optimize stimulation dose and montage and to embed stimulation within task specific gait practice across multiple sessions. Well powered studies with extended follow up should test whether repeated spinal stimulation yields durable gains in posture and locomotion and helps counter age related decline in spinal control.

Concluding Remarks and Perspectives

This thesis investigated the neuromechanical control of walking in older adults by examining how posture, physical activity, and spinal neuromodulation influence gait efficiency and stability. Through a series of experimental chapters, it offered an integrative view that connects mechanical, muscular, and neural adaptations with age-related locomotor changes. More than describing the manifestations of aging gait, this work aimed to uncover the underlying mechanisms that drive these changes and to explore their potential reversibility.

By bridging biomechanics and motor control, the findings demonstrate that aging affects not only the peripheral structures involved in force production but also the timing, coordination, and segmental specificity of spinal motor output. This neuromechanical degradation was found to be partially mitigated by habitual physical activity, which preserved gait coordination and spinal organization even in older adults. Additionally, preliminary tsDCS findings support spinal circuit plasticity, but interindividual variation in subcutaneous tissue thickness at the stimulation site may have reduced current penetration and limited the detectable effects of a single session.

A key insight from this thesis is that gait control in aging cannot be fully understood through isolated parameters. Instead, it must be viewed as an emergent property of a system that integrates posture, movement, and neural regulation in a task-specific manner. Trunk inclination, segmental activation patterns, and transition timing are not peripheral outcomes, but central variables that reflect the broader integrity of locomotor control. Notably, these patterns echo those observed in early development, suggesting a common constraint: reduced corticospinal modulation impairs the efficient redirection of the CoM, whether due to immaturity or degeneration.

Looking ahead, future research should focus on multi-modal interventions that combine exercise with spinal neuromodulation, as well as on long-term studies using objective activity monitoring tools. Special attention should be given to identifying early neuromechanical markers of decline and developing strategies that preserve locomotor function before irreversible deficits emerge. Rather than approaching aging as a linear loss, this thesis supports a perspective in which movement, and the systems that enable it, remain adaptable across the lifespan. The results call for a shift from symptomatic observations toward mechanistic understanding, and from isolated rehabilitation strategies toward integrated, system-wide approaches. In this sense, the study of gait continues to serve as a powerful window into the organization of human movement, and into the ways we might preserve it.

References

Aagaard, P., Suetta, C., Caserotti, P., Magnusson, S. P., & Kjaer, M. (2010). Role of the nervous system in sarcopenia and muscle atrophy with aging: Strength training as a countermeasure. *Scandinavian Journal of Medicine & Science in Sports*, *20*(1), 49–64. https://doi.org/10.1111/j.1600-0838.2009.01084.x

Abell, J., Bennett, G., Cadar, D., Chatzy, G., Crawford, R., Dangerfield, P., Demakakos, P., De Oliveria, C., Gessa, G. D., Hussey, D., Karjalainen, H., Lassale, C., Nazroo, J., Oldfield, Z., Pacchiotti, B., Steptoe, A., Banks, J., & Zaninott, P. (2020). *The Dynamics of Ageing: Evidence from the the English Longitudinal Study of Ageing 2002-2019*.

Abellan Van Kan, G., Rolland, Y., Andrieu, S., Bauer, J., Beauchet, O., Bonnefoy, M., Cesari, M., Donini, L. M., Gillette-Guyonnet, S., Inzitari, M., Nourhashemi, F., Onder, G., Ritz, P., Salva, A., Visser, M., & Vellas, B. (2009). Gait speed at usual pace as a predictor of adverse outcomes in community-dwelling older people an International Academy on Nutrition and Aging (IANA) Task Force. *The Journal of Nutrition, Health and Aging*, *13*(10), 881–889. https://doi.org/10.1007/s12603-009-0246-z

Aboutorabi, A., Arazpour, M., Bahramizadeh, M., Hutchins, S. W., & Fadayevatan, R. (2016). The effect of aging on gait parameters in able-bodied older subjects: A literature review. *Aging Clinical and Experimental Research*, *28*(3), 393–405. https://doi.org/10.1007/s40520-015-0420-6

Adolph, K. E., Hoch, J. E., & Cole, W. G. (2018). Development (of Walking): 15 Suggestions. *Trends in Cognitive Sciences*, *22*(8), 699–711. https://doi.org/10.1016/j.tics.2018.05.010

Adolph, K. E., & Robinson, S. R. (2015). Motor Development. In R. M. Lerner (Ed.), *Handbook of Child Psychology and Developmental Science* (1st ed., pp. 1–45). Wiley. https://doi.org/10.1002/9781118963418.childpsy204

Agyapong-Badu, S., Warner, M., Samuel, D., & Stokes, M. (2016). Measurement of ageing effects on muscle tone and mechanical properties of rectus femoris and biceps brachii in healthy males and females using a novel hand-held myometric device. *Archives of Gerontology and Geriatrics*, *62*, 59–67. https://doi.org/10.1016/j.archger.2015.09.011

Akagi, R., Yamashita, Y., & Ueyasu, Y. (2015). Age-Related Differences in Muscle Shear Moduli in the Lower Extremity. *Ultrasound in*

Medicine & Biology, *41*(11), 2906–2912. https://doi.org/10.1016/j.ultrasmedbio.2015.07.011

Akima, H., Kano, Y., Enomoto, Y., Ishizu, M., Okada, M., Oishi, Y., Katsuta, S., & Kuno, S.-Y. (2001). Muscle function in 164 men and women aged 20-84 yr. *Medicine and Science in Sports and Exercise*, 220–226. https://doi.org/10.1097/00005768-200102000-00008

Alghamdi, W., & Preece, S. J. (2020). How does normal variability in trunk flexion affect lower limb muscle activity during walking? *Human Movement Science*, *72*, 102630. https://doi.org/10.1016/j.humov.2020.102630

Allen, J. L., & Franz, J. R. (2018). The motor repertoire of older adult fallers may constrain their response to balance perturbations. *Journal of Neurophysiology*, *120*(5), 2368–2378. https://doi.org/10.1152/jn.00302.2018

Allen, M. D., Dalton, B. H., Gilmore, K. J., McNeil, C. J., Doherty, T. J., Rice, C. L., & Power, G. A. (2021). Neuroprotective effects of exercise on the aging human neuromuscular system. *Experimental Gerontology*, *152*, 111465. https://doi.org/10.1016/j.exger.2021.111465

Aminiaghdam, S., Blickhan, R., Muller, R., & Rode, C. (2017). Posture alteration as a measure to accommodate uneven ground in able-bodied gait. *PLOS ONE*, *12*(12), e0190135. https://doi.org/10.1371/journal.pone.0190135

Aminiaghdam, S., Müller, R., & Blickhan, R. (2018). Locomotor stability in able-bodied trunk-flexed gait across uneven ground. *Human Movement Science*, *62*, 176–183. https://doi.org/10.1016/j.humov.2018.10.011

Aminiaghdam, S., Rode, C., Müller, R., & Blickhan, R. (2016). Increasing trunk flexion transforms human leg function into that of birds despite different leg morphology. *Journal of Experimental Biology*, jeb.148312. https://doi.org/10.1242/jeb.148312

Andersen, J. L. (2003). Muscle fibre type adaptation in the elderly human muscle. *Scandinavian Journal of Medicine & Science in Sports*, *13*(1), 40–47. https://doi.org/10.1034/j.1600-0838.2003.00299.x

Andre-Thomas, Y., & Autgaerden, S. (1966). *Locomotion from Pre- to Post-Natal Life: How the Newborn Begins to Acquire Psycho-Sensory Functions* (First). https://shop.elsevier.com/books/locomotion-from-pre-to-post-natal-life/thomas/978-1-4831-9739-5

Angeli, C. A., Boakye, M., Morton, R. A., Vogt, J., Benton, K., Chen, Y., Ferreira, C. K., & Harkema, S. J. (2018). Recovery of Over-Ground Walking after Chronic Motor Complete Spinal Cord Injury. *New England Journal of Medicine*, *379*(13), 1244–1250. https://doi.org/10.1056/NEJMoa1803588

Angius, L., Hopker, J., & Mauger, A. R. (2017). The Ergogenic Effects of Transcranial Direct Current Stimulation on Exercise Performance. *Frontiers in Physiology*, *8*. https://doi.org/10.3389/fphys.2017.00090

Armitano-Lago, C., Pietrosimone, B., Davis-Wilson, H. C., Evans-Pickett, A., Franz, J. R., Blackburn, T., & Kiefer, A. W. (2020). Biofeedback augmenting lower limb loading alters the underlying temporal structure of gait following anterior cruciate ligament reconstruction. *Human Movement Science*, *73*, 102685. https://doi.org/10.1016/j.humov.2020.102685

Avaltroni, P., Cappellini, G., Sylos-Labini, F., Ivanenko, Y., & Lacquaniti, F. (2024). Spinal maps of motoneuron activity during human locomotion: Neuromechanical considerations. *Frontiers in Physiology*, *15*, 1389436. https://doi.org/10.3389/fphys.2024.1389436

Awosika, O. O., Sandrini, M., Volochayev, R., Thompson, R. M., Fishman, N., Wu, T., Floeter, M. K., Hallett, M., & Cohen, L. G. (2019). Transcutaneous spinal direct current stimulation improves locomotor learning in healthy humans. *Brain Stimulation*, *12*(3), 628–634. https://doi.org/10.1016/j.brs.2019.01.017

Bach, M. M., Daffertshofer, A., & Dominici, N. (2021). The development of mature gait patterns in children during walking and running. *European Journal of Applied Physiology*, *121*(4), 1073–1085. https://doi.org/10.1007/s00421-020-04592-2

Backman, E., Odenrick, P., Henriksson, K. G., & Ledin, T. (1989). Isometric muscle force and anthropometric values in normal children aged between 3.5 and 15 years. *Journal of Rehabilitation Medicine*, *21*(2), 105–114. https://doi.org/10.2340/165019771989105114

Baczyk, M., D.-C. H., Mrówczynski W. ,. Krutki, P. (2019). Motoneuron firing properties are modified by trans-spinal direct current stimulation in rats. *Journal of Applied Physiology*, *126*(5), 1232–1241. https://doi.org/10.1152/japplphysiol.00803.2018

Barliya, A., Omlor, L., Giese, M. A., & Flash, T. (2009). An analytical formulation of the law of intersegmental coordination during human locomotion. *Experimental Brain Research*, *193*(3), 371–385. https://doi.org/10.1007/s00221-008-1633-0

Baş, H., Okyar Baş, A., Ceylan, S., Güner, M., Koca, M., Hafızoğlu, M., Şahiner, Z., Öztürk, Y., Balcı, C., Doğu, B. B., Cankurtaran, M., & Halil, M. G. (2023). Lower gastrocnemius muscle stiffness, derived from elastography, is an independent factor for falls in older adults. *Aging Clinical and Experimental Research*, *35*(12), 2979–2986. https://doi.org/10.1007/s40520-023-02605-6

Bastien, G. J., Gosseye, T. P., & Penta, M. (2019). A robust machine learning enabled decomposition of shear ground reaction forces during the double contact phase of walking. *Gait & Posture*, *73*, 221–227. https://doi.org/10.1016/j.gaitpost.2019.07.190

Bastien, G. J., Heglund, N. C., & Schepens, B. (2003). The double contact phase in walking children. *Journal of Experimental Biology*, *206*(17), 2967–2978. https://doi.org/10.1242/jeb.00494

Batschelet, E. (1981). *Circular Statistics in Biology*. Academic Press.

Batsis, J. A., Mackenzie, T. A., Barre, L. K., Lopez-Jimenez, F., & Bartels, S. J. (2014). Sarcopenia, sarcopenic obesity and mortality in older adults: Results from the National Health and Nutrition Examination

Survey III. *European Journal of Clinical Nutrition*, *68*(9), 1001–1007. https://doi.org/10.1038/ejcn.2014.117

Baudry, S., Collignon, S., & Duchateau, J. (2015). Influence of age and posture on spinal and corticospinal excitability. *Experimental Gerontology*, *69*, 62–69. https://doi.org/10.1016/j.exger.2015.06.006

Bean, J. F., Leveille, S. G., Kiely, D. K., Bandinelli, S., Guralnik, J. M., & Ferrucci, L. (2003). A Comparison of Leg Power and Leg Strength Within the InCHIANTI Study: Which Influences Mobility More? *The Journals of Gerontology Series A: Biological Sciences and Medical Sciences*, *58*(8), M728–M733. https://doi.org/10.1093/gerona/58.8.M728

Beaudart, C., Zaaria, M., Pasleau, F., Reginster, J.-Y., & Bruyère, O. (2017). Health Outcomes of Sarcopenia: A Systematic Review and Meta-Analysis. *PLOS ONE*, *12*(1), e0169548. https://doi.org/10.1371/journal.pone.0169548

Beck, O. N., Kipp, S., Roby, J. M., Grabowski, A. M., Kram, R., & Ortega, J. D. (2016). Older runners retain youthful running economy despite biomechanical differences. *Medicine and Science in Sports and Exercise*, *48*(4), 697–704. Scopus. https://doi.org/10.1249/MSS.0000000000000820

Beijersbergen, C. M. I., Granacher, U., Gäbler, M., DeVita, P., & Hortobágyi, T. (2017). Hip mechanics underlie lower extremity power training-induced increase in old adults' fast gait velocity: The Potsdam Gait Study (POGS). *Gait & Posture*, *52*, 338–344. https://doi.org/10.1016/j.gaitpost.2016.12.024

Beijersbergen, C. M. I., Granacher, U., Gäbler, M., Devita, P., & Hortobágyi, T. (2017a). Kinematic Mechanisms of How Power Training Improves Healthy Old Adults' Gait Velocity. *Medicine & Science in Sports & Exercise*, *49*(1), 150–157. https://doi.org/10.1249/MSS.0000000000001082

Beijersbergen, C. M. I., Granacher, U., Gäbler, M., Devita, P., & Hortobágyi, T. (2017b). Power Training–induced Increases in Muscle Activation during Gait in Old Adults. *Medicine & Science in Sports & Exercise*, *49*(11), 2198–2025. https://doi.org/10.1249/MSS.0000000000001345

Beijersbergen, C. M. I., Granacher, U., Vandervoort, A. A., DeVita, P., & Hortobágyi, T. (2013). The biomechanical mechanism of how strength and power training improves walking speed in old adults remains unknown. *Ageing Research Reviews*, *12*(2), 618–627. https://doi.org/10.1016/j.arr.2013.03.001

Bernard, J. A., & Seidler, R. D. (2014). Moving forward: Age effects on the cerebellum underlie cognitive and motor declines. *Neuroscience & Biobehavioral Reviews*, *42*, 193–207. https://doi.org/10.1016/j.neubiorev.2014.02.011

Berry, H. R., Tate, R. J., & Conway, B. A. (2017). Transcutaneous spinal direct current stimulation induces lasting fatigue resistance and enhances explosive vertical jump performance. *PLOS ONE*, *12*(4), e0173846. https://doi.org/10.1371/journal.pone.0173846

Bertsch, C., Unger, H., Winkelmann, W., & Rosenbaum, D. (2004). Evaluation of early walking patterns from plantar pressure distribution measurements. First year results of 42 children. *Gait & Posture*, *19*(3), 235–242. https://doi.org/10.1016/S0966-6362(03)00064-X

Bhargava, L. J., Pandy, M. G., & Anderson, F. C. (2004). A phenomenological model for estimating metabolic energy consumption in muscle contraction. *Journal of Biomechanics*, *37*(1), 81–88. https://doi.org/10.1016/S0021-9290(03)00239-2

Bianchi, L., Angelini, D., & Lacquaniti, F. (1998). Individual characteristics of human walking mechanics. *Pfl□gers Archiv European Journal of*

Physiology, *436*(3), 343–356. https://doi.org/10.1007/s004240050642

Bianchi, L., Angelini, D., Orani, G. P., & Lacquaniti, F. (1998). Kinematic Coordination in Human Gait: Relation to Mechanical Energy Cost. *Journal of Neurophysiology*, *79*(4), 2155–2170. https://doi.org/10.1152/jn.1998.79.4.2155

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

Bleyenheuft, C., & Detrembleur, C. (2012). Kinematic covariation in pediatric, adult and elderly subjects: Is gait control influenced by age? *Clinical Biomechanics*, *27*(6), 568–572. https://doi.org/10.1016/j.clinbiomech.2012.01.010

Bocci, T., Barloscio, D., Vergari, M., Di Rollo, A., Rossi, S., Priori, A., & Sartucci, F. (2015). Spinal Direct Current Stimulation Modulates Short Intracortical Inhibition. *Neuromodulation: Technology at the Neural Interface*, *18*(8), 686–693. https://doi.org/10.1111/ner.12298

Borghese, N. A., Bianchi, L., & Lacquaniti, F. (1996). Kinematic determinants of human locomotion. *The Journal of Physiology*, *494*(3), 863–879. https://doi.org/10.1113/jphysiol.1996.sp021539

Bornstein, M. H., & Bornstein, H. G. (1976). The pace of life. *Nature*, *259*, 557–559. https://doi.org/10.1038/259557a0

Borzuola, R., Giombini, A., Torre, G., Campi, S., Albo, E., Bravi, M., Borrione, P., Fossati, C., & Macaluso, A. (2020). Central and Peripheral Neuromuscular Adaptations to Ageing. *Journal of Clinical Medicine*, *9*(3), 741. https://doi.org/10.3390/jcm9030741

Bosch, K., Gerss, J., & Rosenbaum, D. (2007). Preliminary normative values for foot loading parameters of the developing child. *Gait & Posture*, *26*(2), 238–247. https://doi.org/10.1016/j.gaitpost.2006.09.014

Boyer, K. A., Andriacchi, T. P., & Beaupre, G. S. (2012). The role of physical activity in changes in walking mechanics with age. *Gait & Posture*, *36*(1), 149–153. https://doi.org/10.1016/j.gaitpost.2012.02.007

Boyer, K. A., Hayes, K. L., Umberger, B. R., Adamczyk, P. G., Bean, J. F., Brach, J. S., Clark, B. C., Clark, D. J., Ferrucci, L., Finley, J., Franz, J. R., Golightly, Y. M., Hortobágyi, T., Hunter, S., Narici, M., Nicklas, B., Roberts, T., Sawicki, G., Simonsick, E., & Kent, J. A. (2023). Age-related changes in gait biomechanics and their impact on the metabolic cost of walking: Report from a National Institute on Aging workshop. *Experimental Gerontology*, *173*, 112102. https://doi.org/10.1016/j.exger.2023.112102

Boyer, K. A., Johnson, R. T., Banks, J. J., Jewell, C., & Hafer, J. F. (2017). Systematic review and meta-analysis of gait mechanics in young and older adults. *Experimental Gerontology*, *95*, 63–70. https://doi.org/10.1016/j.exger.2017.05.005

Brach, J. S., Van Swearingen, J. M., Perera, S., Wert, D. M., & Studenski, S. (2013). Motor learning versus standard walking exercise in older adults with subclinical gait dysfunction: A randomized clinical trial. *Journal of the American Geriatrics Society*, *61*(11), 1879–1886. Scopus. https://doi.org/10.1111/jgs.12506

Brenière, Y., & Bril, B. (1998). Development of postural control of gravity forces in children during the first 5 years of walking. *Experimental Brain Research*, *121*(3), 255–262. https://doi.org/10.1007/s002210050458

Browne, M. G., & Franz, J. R. (2019). Ankle power biofeedback attenuates the distal-to-proximal redistribution in older adults. *Gait & Posture*, *71*, 44–49. https://doi.org/10.1016/j.gaitpost.2019.04.011

Bua, E., Johnson, J., Herbst, A., Delong, B., McKenzie, D., Salamat, S., & Aiken, J. M. (2006). Mitochondrial DNA–Deletion Mutations Accumulate Intracellularly to Detrimental Levels in Aged Human Skeletal Muscle Fibers. *The American Journal of Human Genetics*, *79*(3), 469–480. https://doi.org/10.1086/507132

Bull, F. C., Al-Ansari, S. S., Biddle, S., Borodulin, K., Buman, M. P., Cardon, G., Carty, C., Chaput, J.-P., Chastin, S., Chou, R., Dempsey, P. C., DiPietro, L., Ekelund, U., Firth, J., Friedenreich, C. M., Garcia, L., Gichu, M., Jago, R., Katzmarzyk, P. T., … Willumsen, J. F. (2020). World Health Organization 2020 guidelines on physical activity and sedentary behaviour. *British Journal of Sports Medicine*, *54*(24), 1451–1462. https://doi.org/10.1136/bjsports-2020-102955

Cade, W. T., & Yarasheski, K. E. (2006). Metabolic and Molecular Aspects of Sarcopenia. In *Principles of Molecular Medicine* (pp. 529–534).

Cappellini, G., Ivanenko, Y. P., Dominici, N., Poppele, R. E., & Lacquaniti, F. (2010). Migration of Motor Pool Activity in the Spinal Cord Reflects Body Mechanics in Human Locomotion. *Journal of*

Neurophysiology, *104*(6), 3064–3073. https://doi.org/10.1152/jn.00318.2010

Cavagna, G. A., Franzetti, P., & Fuchimoto, T. (1983). The mechanics of walking in children. *The Journal of Physiology*, *343*(1), 323–339. https://doi.org/10.1113/jphysiol.1983.sp014895

Cavagna, G. A., Heglund, N. C., & Taylor, C. R. (1977). Mechanical work in terrestrial locomotion: Two basic mechanisms for minimizing energy expenditure. *American Journal of Physiology-Regulatory, Integrative and Comparative Physiology*, *233*(5), R243–R261. https://doi.org/10.1152/ajpregu.1977.233.5.R243

Cavagna, G. A., Thys, H., & Zamboni, A. (1976). The sources of external work in level walking and running. *The Journal of Physiology*, *262*(3), 639–657. https://doi.org/10.1113/jphysiol.1976.sp011613

Çekok, F. K., Taş, S., & Aktaş, A. (2024). Muscle and tendon stiffness of lower extremity in older adults with fall history: Stiffness effect on physical performance and fall risk. *Geriatric Nursing*, *59*, 228–233. https://doi.org/10.1016/j.gerinurse.2024.07.003

Chastan, N., Westby, G. W. M., Yelnik, J., Bardinet, E., Do, M. C., Agid, Y., & Welter, M. L. (2008). Effects of nigral stimulation on locomotion and postural stability in patients with Parkinson's disease. *Brain*, *132*(1), 172–184. https://doi.org/10.1093/brain/awn294

Chen, N., He, X., Feng, Y., Ainsworth, B. E., & Liu, Y. (2021). Effects of resistance training in healthy older people with sarcopenia: A systematic review and meta-analysis of randomized controlled trials. *European Review of Aging and Physical Activity*, *18*(1), 23. https://doi.org/10.1186/s11556-021-00277-7

Cheron, G., Bengoetxea, A., Bouillot, E., Lacquaniti, F., & Dan, B. (2001). Early emergence of temporal co-ordination of lower limb segments elevation angles in human locomotion. *Neuroscience Letters*, *308*(2), 123–127. https://doi.org/10.1016/S0304-3940(01)01925-5

Cheron, G., Bouillot, E., Dan, B., Bengoetxea, A., Draye, J.-P., & Lacquaniti, F. (2001). Development of a kinematic coordination pattern in toddler locomotion: Planar covariation. *Experimental Brain Research*, *137*(3–4), 455–466. https://doi.org/10.1007/s002210000663

Christie, J. (2011). Progressive resistance strength training for improving physical function in older adults. *International Journal of Older People Nursing*, *6*(3), 244–246. https://doi.org/10.1111/j.1748-3743.2011.00291.x

Clark, B. C., & Manini, T. M. (2008). Sarcopenia = Dynapenia. *The Journals of Gerontology Series A: Biological Sciences and Medical Sciences*, *63*(8), 829–834. https://doi.org/10.1093/gerona/63.8.829

Cogliati, M., Cudicio, A., Martinez-Valdes, E., Tarperi, C., Schena, F., Orizio, C., & Negro, F. (2020). Half marathon induces changes in central control and peripheral properties of individual motor units in master athletes. *Journal of Electromyography and Kinesiology*, *55*. Scopus. https://doi.org/10.1016/j.jelekin.2020.102472

Cooper, R., Kuh, D., Cooper, C., Gale, C. R., Lawlor, D. A., Matthews, F., Hardy, R., & the FALCon and HALCyon Study Teams. (2011). Objective measures of physical capability and subsequent health: A systematic review. *Age and Ageing*, *40*(1), 14–23. https://doi.org/10.1093/ageing/afq117

Cooper, R., Kuh, D., Hardy, R., & Group, M. R. (2010). Objectively measured physical capability levels and mortality: Systematic review and meta-analysis. *BMJ*, *341*, c4467. https://doi.org/10.1136/bmj.c4467

Coscia, M., Monaco, V., Martelloni, C., Rossi, B., Chisari, C., & Micera, S. (2015). Muscle synergies and spinal maps are sensitive to the asymmetry induced by a unilateral stroke. *Journal of*

NeuroEngineering and Rehabilitation, *12*(1), 39. https://doi.org/10.1186/s12984-015-0031-7

Crompton, R. H., Weijie, L. Y. W., Günther, M., & Savage, R. (1998). The mechanical effectiveness of erect and “bent-hip, bent-knee” bipedal walking inAustralopithecus afarensis. *Journal of Human Evolution*, *35*(1), 55–74. https://doi.org/10.1006/jhev.1998.0222

Cruz-Jentoft, A. J., Bahat, G., Bauer, J., Boirie, Y., Bruyère, O., Cederholm, T., Cooper, C., Landi, F., Rolland, Y., Sayer, A. A., Schneider, S. M., Sieber, C. C., Topinkova, E., Vandewoude, M., Visser, M., Zamboni, M., Writing Group for the European Working Group on Sarcopenia in Older People 2 (EWGSOP2), and the Extended Group for EWGSOP2, Bautmans, I., Baeyens, J.-P., … Schols, J. (2019). Sarcopenia: Revised European consensus on definition and diagnosis. *Age and Ageing*, *48*(1), 16–31. https://doi.org/10.1093/ageing/afy169

da Cunha, J., Maselli, L. M. F., Stern, A. C. B., Spada, C., & Bydlowski, S. P. (2015). Impact of antiretroviral therapy on lipid metabolism of human immunodeficiency virus-infected patients: Old and new drugs. *World Journal of Virology*, *4*(2), 56–77. https://doi.org/10.5501/wjv.v4.i2.56

Damasceno, A., Delicio, A. M., Mazo, D. F. C., Zullo, J. F. D., Scherer, P., T.Y. Ng, R., & Damasceno, B. P. (2005). Primitive reflexes and cognitive function. *Arquivos de Neuro-Psiquiatria*, *63*(3a), 577–582. https://doi.org/10.1590/S0004-282X2005000400004

Das Gupta, S., Bobbert, M. F., & Kistemaker, D. A. (2019). The Metabolic Cost of Walking in healthy young and older adults – A Systematic Review and Meta Analysis. *Scientific Reports*, *9*(1), 9956. https://doi.org/10.1038/s41598-019-45602-4

Davis, S. W., Dennis, N. A., Buchler, N. G., White, L. E., Madden, D. J., & Cabeza, R. (2009). Assessing the effects of age on long white matter tracts using diffusion tensor tractography. *NeuroImage*, *46*(2), 530–541. https://doi.org/10.1016/j.neuroimage.2009.01.068

Dayanidhi, S., Kutch, J. J., & Valero-Cuevas, F. J. (2013). Decrease in Muscle Contraction Time Complements Neural Maturation in the Development of Dynamic Manipulation. *The Journal of Neuroscience*, *33*(38), 15050–15055. https://doi.org/10.1523/JNEUROSCI.1968-13.2013

DeJaeger, D., Willems, P. A., & Heglund, N. C. (2001). The energy cost of walking in children. *European Journal of Physiology*, *441*(4), 538–543. https://doi.org/10.1007/s004240000443

Delabastita, T., Hollville, E., Catteau, A., Cortvriendt, P., De Groote, F., & Vanwanseele, B. (2021). Distal-to-proximal joint mechanics redistribution is a main contributor to reduced walking economy in older adults. *Scandinavian Journal of Medicine & Science in Sports*, *31*(5), 1036–1047. https://doi.org/10.1111/sms.13929

Delmonico, M. J., Harris, T. B., Visser, M., Park, S. W., Conroy, M. B., Velasquez-Mieyer, P., Boudreau, R., Manini, T. M., Nevitt, M., Newman, A. B., & Goodpaster, B. H. (2009). Longitudinal study of muscle strength, quality, and adipose tissue infiltration. *The American Journal of Clinical Nutrition*, *90*(6), 1579–1585. https://doi.org/10.3945/ajcn.2009.28047

Delsys. (2023). *Trigno Wireless Biofeedback System User's Guide* (p. 35). https://delsys.com/downloads/USERSGUIDE/MAN-031-1-7%20Trigno%20Wireless%20Biofeedback%20System.pdf

Demes, B., Thompson, N. E., O'Neill, M. C., & Umberger, B. R. (2015). Center of mass mechanics of chimpanzee bipedal walking. *American Journal of Physical Anthropology*, *156*(3), 422–433. https://doi.org/10.1002/ajpa.22667

Dempster, W. T. (1955). The Anthropometry of Body Action. *Annals of the New York Academy of Sciences*, *63*(4), 559–585. https://doi.org/10.1111/j.1749-6632.1955.tb32112.x

Deschenes, M. R., Roby, M. A., Eason, M. K., & Harris, M. B. (2010). Remodeling of the neuromuscular junction precedes sarcopenia related alterations in myofibers. *Experimental Gerontology*, *45*(5), 389–393. https://doi.org/10.1016/j.exger.2010.03.007

DeVita, P., & Hortobagyi, T. (2000). Age causes a redistribution of joint torques and powers during gait. *Journal of Applied Physiology*, *88*(5), 1804–1811. Scopus.

Dewolf, A. H., Ivanenko, Y. P., Lacquaniti, F., & Willems, P. A. (2017). Pendular energy transduction within the step during human walking on slopes at different speeds. *PLOS ONE*, *12*(10), e0186963. https://doi.org/10.1371/journal.pone.0186963

Dewolf, A. H., Ivanenko, Y. P., Mesquita, R. M., & Willems, P. A. (2021). Postural control in the elephant. *Journal of Experimental Biology*, *224*(22), jeb243648. https://doi.org/10.1242/jeb.243648

Dewolf, A. H., Ivanenko, Y. P., Zelik, K. E., Lacquaniti, F., & Willems, P. A. (2019a). Differential activation of lumbar and sacral motor pools during walking at different speeds and slopes. *Journal of*

Neurophysiology, *122*(2), 872–887. https://doi.org/10.1152/jn.00167.2019

Dewolf, A. H., Ivanenko, Y., Zelik, K. E., Lacquaniti, F., & Willems, P. A. (2018). Kinematic patterns while walking on a slope at different speeds. *Journal of Applied Physiology*, *125*(2), 642–653. https://doi.org/10.1152/japplphysiol.01020.2017

Dewolf, A. H., Meurisse, G. M., Ivanenko, Y., Lacquaniti, F., Bastien, G. J., & Schepens, B. (2022a). Relation between Step-To-Step Transition Strategies and Walking Pattern in Older Adults. *Applied Sciences*, *12*(10), 5055. https://doi.org/10.3390/app12105055

Dewolf, A. H., Meurisse, G. M., Schepens, B., & Willems, P. A. (2019). Effect of walking speed on the intersegmental coordination of lower-limb segments in elderly adults. *Gait & Posture*, *70*, 156–161. https://doi.org/10.1016/j.gaitpost.2019.03.001

Dewolf, A. H., Peñailillo, L. E., & Willems, P. A. (2016a). The rebound of the body during uphill and downhill running at different speeds. *Journal of Experimental Biology*, jeb.142976. https://doi.org/10.1242/jeb.142976

Dewolf, A. H., Sylos-Labini, F., Cappellini, G., Lacquaniti, F., & Ivanenko, Y. (2020). Emergence of Different Gaits in Infancy: Relationship

Between Developing Neural Circuitries and Changing Biomechanics. *Frontiers in Bioengineering and Biotechnology*, *8*. https://doi.org/10.3389/fbioe.2020.00473

Dewolf, A. H., Sylos-Labini, F., Cappellini, G., Zhvansky, D., Willems, P. A., Ivanenko, Y., & Lacquaniti, F. (2021). Neuromuscular Age-Related Adjustment of Gait When Moving Upwards and Downwards. *Frontiers in Human Neuroscience*, *15*, 749366. https://doi.org/10.3389/fnhum.2021.749366

Dewolf, A. H., & Willems, P. A. (2019). Running on a slope: A collision-based analysis to assess the optimal slope. *Journal of Biomechanics*, *83*, 298–304. https://doi.org/10.1016/j.jbiomech.2018.12.024

Doherty, T. J. (2003). Invited Review: Aging and sarcopenia. *Journal of Applied Physiology*, *95*(4), 1717–1727. https://doi.org/10.1152/japplphysiol.00347.2003

Doherty, T. J., Vandervoort, A. A., Taylor, A. W., & Brown, W. F. (1993). Effects of motor unit losses on strength in older men and women. *Journal of Applied Physiology*, *74*(2), 868–874. https://doi.org/10.1152/jappl.1993.74.2.868

Domingues-Faria, C., Vasson, M.-P., Goncalves-Mendes, N., Boirie, I., & Walrand, S. (2016). Skeletal muscle regeneration and impact of aging and nutrition. *Ageing Research Reviews*, *26*, 22–36.

Dominici, N., Ivanenko, Y. P., Cappellini, G., d'Avella, A., Mondì, V., Cicchese, M., Fabiano, A., Silei, T., Di Paolo, A., Giannini, C., Poppele, R. E., & Lacquaniti, F. (2011). Locomotor Primitives in Newborn Babies and Their Development. *Science*, *334*(6058), 997–999. https://doi.org/10.1126/science.1210617

Dominici, N., Ivanenko, Y. P., Cappellini, G., Zampagni, M. L., & Lacquaniti, F. (2010). Kinematic Strategies in Newly Walking Toddlers Stepping Over Different Support Surfaces. *Journal of Neurophysiology*, *103*(3), 1673–1684. https://doi.org/10.1152/jn.00945.2009

Dominici, N., Ivanenko, Y. P., & Lacquaniti, F. (2007). Control of Foot Trajectory in Walking Toddlers: Adaptation to Load Changes. *Journal of Neurophysiology*, *97*(4), 2790–2801. https://doi.org/10.1152/jn.00262.2006

Donelan, J. M., Kram, R., & Kuo, A. D. (2002). Simultaneous positive and negative external mechanical work in human walking. *Journal of*

Biomechanics, *35*(1), 117–124. https://doi.org/10.1016/S0021-9290(01)00169-5

Douglas Gossman, M., & Lawrence Jacobs, M. D. (1980). Three primitive reflexes in parkinsonism patients. *Neurology*, *30*(2), 189.

Dragoi, I. I., Popescu, F. G., Petrita, T., Alexa, F., Barac, S., Bondor, C. I., Pauncu, E.-A., Bowling, F. L., Reeves, N. D., & Ionac, M. (2022). Acute Effects of Sedentary Behavior on Ankle Torque Assessed with a Custom-Made Electronic Dynamometer. *Journal of Clinical Medicine*, *11*(9), 2474. https://doi.org/10.3390/jcm11092474

Eberhardt, F., Enax-Krumova, E., Tegenthoff, M., Höffken, O., & Özgül, Ö. S. (2023). Anodal transcutaneous spinal direct current stimulation influences the amplitude of pain-related evoked potentials in healthy subjects. *Scientific Reports*, *13*(1). https://doi.org/10.1038/s41598-023-47408-x

el Hadouchi, M., Kiers, H., de Vries, R., Veenhof, C., & van Dieën, J. (2022). Effectiveness of power training compared to strength training in older adults: A systematic review and meta-analysis. *European Review of Aging and Physical Activity*, *19*(1), 1–15. https://doi.org/10.1186/s11556-022-00297-x

Fava De Lima, F., Silva, C. R., & Kohn, A. F. (2022). Transcutaneous spinal direct current stimulation (tsDCS) does not affect postural sway of young and healthy subjects during quiet upright standing. *PLOS ONE*, *17*(4), e0267718. https://doi.org/10.1371/journal.pone.0267718

Feltes, B. C., De Faria Poloni, J., & Bonatto, D. (2015). Development and Aging: Two Opposite but Complementary Phenomena. In A. I. Yashin & S. M. Jazwinski (Eds.), *Interdisciplinary Topics in Gerontology and Geriatrics* (Vol. 40, pp. 74–84). S. Karger AG. https://doi.org/10.1159/000364932

Fernandes, S. R., Salvador, R., Wenger, C., De Carvalho, M., & Miranda, P. C. (2018). Transcutaneous spinal direct current stimulation of the lumbar and sacral spinal cord: A modelling study. *Journal of Neural Engineering*, *15*(3), 036008. https://doi.org/10.1088/1741-2552/aaac38

Fiatarone, M. A. (1990). High-Intensity Strength Training in Nonagenarians: Effects on Skeletal Muscle. *JAMA*, *263*(22), 3029. https://doi.org/10.1001/jama.1990.03440220053029

Fielding, R. A., Vellas, B., Evans, W. J., Bhasin, S., Morley, J. E., Newman, A. B., Abellan Van Kan, G., Andrieu, S., Bauer, J., Breuille, D., Cederholm, T., Chandler, J., De Meynard, C., Donini, L., Harris, T., Kannt, A., Keime Guibert, F., Onder, G., Papanicolaou, D., … Zamboni, M. (2011). Sarcopenia: An Undiagnosed Condition in Older Adults. Current Consensus Definition: Prevalence, Etiology, and Consequences. International Working Group on Sarcopenia. *Journal of the American Medical Directors Association*, *12*(4), 249–256. https://doi.org/10.1016/j.jamda.2011.01.003

Forssberg, H. (1985). Ontogeny of human locomotor control I. Infant stepping, supported locomotion and transition to independent locomotion. *Experimental Brain Research*, *57*(3). https://doi.org/10.1007/BF00237835

Foster, R. J., Maganaris, C. N., Reeves, N. D., & Buckley, J. G. (2019). Centre of mass control is reduced in older people when descending stairs at an increased riser height. *Gait & Posture*, *73*, 305–314. https://doi.org/10.1016/j.gaitpost.2019.08.004

Franz, J. R. (2016). The Age-Associated Reduction in Propulsive Power Generation in Walking. *Exercise and Sport Sciences Reviews*, *44*(4), 129–136. https://doi.org/10.1249/JES.0000000000000086

Franz, J. R., & Kram, R. (2013). How does age affect leg muscle activity/coactivity during uphill and downhill walking? *Gait &*

Posture, *37*(3), 378–384. https://doi.org/10.1016/j.gaitpost.2012.08.004

Freiberger, E., Sieber, C. C., & Kob, R. (2020). Mobility in Older Community-Dwelling Persons: A Narrative Review. *Frontiers in Physiology*, *11*, 881. https://doi.org/10.3389/fphys.2020.00881

Full, R. J., & Koditschek, D. E. (1999). Templates and anchors: Neuromechanical hypotheses of legged locomotion on land. *Journal of Experimental Biology*, *202*(23), 3325–3332.

Gad, P., Hastings, S., Zhong, H., Seth, G., Kandhari, S., & Edgerton, V. R. (2021). Transcutaneous Spinal Neuromodulation Reorganizes Neural Networks in Patients with Cerebral Palsy. *Neurotherapeutics*, *18*(3), 1953–1962. https://doi.org/10.1007/s13311-021-01087-6

Garcia-Bernal, M.-I., Heredia-Rizo, A. M., Gonzalez-Garcia, P., Cortés-Vega, M.-D., & Casuso-Holgado, M. J. (2021a). Validity and reliability of myotonometry for assessing muscle viscoelastic properties in patients with stroke: A systematic review and meta-analysis. *Scientific Reports*, *11*(1). https://doi.org/10.1038/s41598-021-84656-1

Garcia-Bernal, M.-I., Heredia-Rizo, A. M., Gonzalez-Garcia, P., Cortés-Vega, M.-D., & Casuso-Holgado, M. J. (2021b). Validity and reliability of myotonometry for assessing muscle viscoelastic properties in patients with stroke: A systematic review and meta-analysis. *Scientific Reports*, *11*(1), 5062. https://doi.org/10.1038/s41598-021-84656-1

Gava, P., Kern, H., & Carraro, U. (2015). Age-Associated Power Decline from Running, Jumping, and Throwing Male Masters World Records. *Experimental Aging Research*, *41*(2), 115–135. https://doi.org/10.1080/0361073X.2015.1001648

Geirsdottir, O. G., Chang, M., Briem, K., Jonsson, P. V., Thorsdottir, I., & Ramel, A. (2022). Determinants of physical function in community dwelling old people. *Journal of Gerontology and Geriatrics*, *70*(1), 9–16. https://doi.org/10.36150/2499-6564-N261

Gerasimenko, Y., Gorodnichev, R., Moshonkina, T., Sayenko, D., Gad, P., & Reggie Edgerton, V. (2015). Transcutaneous electrical spinal-cord stimulation in humans. *Annals of Physical and Rehabilitation Medicine*, *58*(4), 225–231. https://doi.org/10.1016/j.rehab.2015.05.003

Gollhofer, A., Strojnik, V., Rapp, W., & Schweizer, L. (1992). Behaviour of triceps surae muscle-tendon complex in different jump conditions. *European Journal of Applied Physiology and Occupational*

Physiology, *64*(4), 283–291. Scopus. https://doi.org/10.1007/BF00636213

Good, C. D., Johnsrude, I. S., Ashburner, J., Henson, R. N. A., Friston, K. J., & Frackowiak, R. S. J. (2001). A Voxel-Based Morphometric Study of Ageing in 465 Normal Adult Human Brains. *NeuroImage*, *14*(1), 21–36. https://doi.org/10.1006/nimg.2001.0786

Goodpaster, B. H., Park, S. W., Harris, T. B., Kritchevsky, S. B., Nevitt, M., Schwartz, A. V., Simonsick, E. M., Tylavsky, F. A., Visser, M., Newman, A. B., & for the Health ABC Study. (2006). The Loss of Skeletal Muscle Strength, Mass, and Quality in Older Adults: The Health, Aging and Body Composition Study. *The Journals of Gerontology Series A: Biological Sciences and Medical Sciences*, *61*(10), 1059–1064. https://doi.org/10.1093/gerona/61.10.1059

Gould, N., Moreland, M., Alvarez, R., Trevino, S., & Fenwick, J. (1989). Development of the Child's Arch. *Foot & Ankle*, *9*(5), 241–245. https://doi.org/10.1177/107110078900900506

Graça, A. L., Gomez-Florit, M., Gomes, M. E., & Docheva, D. (2023). Tendon Aging. *Sub-Cellular Biochemistry*, *103*, 121–147. https://doi.org/10.1007/978-3-031-26576-1_7

Grasso, R. (2004). Distributed plasticity of locomotor pattern generators in spinal cord injured patients. *Brain*, *127*(5), 1019–1034. https://doi.org/10.1093/brain/awh115

Grasso, R., Peppe, A., Stratta, F., Angelini, D., Zago, M., Stanzione, P., & Lacquaniti, F. (1999). Basal ganglia and gait control: Apomorphine administration and internal pallidum stimulation in Parkinson's disease. *Experimental Brain Research*, *126*(2), 139–148. https://doi.org/10.1007/s002210050724

Grasso, R., Zago, M., & Lacquaniti, F. (2000). Interactions Between Posture and Locomotion: Motor Patterns in Humans Walking With Bent Posture Versus Erect Posture. *Journal of Neurophysiology*, *83*(1), 288–300. https://doi.org/10.1152/jn.2000.83.1.288

Gueugnon, M., Stapley, P. J., Gouteron, A., Lecland, C., Morisset, C., Casillas, J.-M., Ornetti, P., & Laroche, D. (2019b). Age-Related Adaptations of Lower Limb Intersegmental Coordination During Walking. *Frontiers in Bioengineering and Biotechnology*, *7*, 173. https://doi.org/10.3389/fbioe.2019.00173

Guralnik, J. M., Ferrucci, L., Pieper, C. F., Leveille, S. G., Markides, K. S., Ostir, G. V., Studenski, S., Berkman, L. F., & Wallace, R. B. (2000). Lower Extremity Function and Subsequent Disability: Consistency

Across Studies, Predictive Models, and Value of Gait Speed Alone Compared With the Short Physical Performance Battery. *The Journals of Gerontology Series A: Biological Sciences and Medical Sciences*, *55*(4), M221–M231. https://doi.org/10.1093/gerona/55.4.M221

Guralnik, J. M., Ferrucci, L., Simonsick, E. M., Salive, M. E., & Wallace, R. B. (1995). Lower-extremity function in persons over the age of 70 years as a predictor of subsequent disability. *The New England Journal of Medicine*, *332*(9), 556–561. https://doi.org/10.1056/NEJM199503023320902

Hallemans, A., De Clercq, D., Dongen, S. V., & Aerts, P. (2006). Changes in foot-function parameters during the first 5 months after the onset of independent walking: A longitudinal follow-up study. *Gait & Posture*, *23*(2), 142–148. https://doi.org/10.1016/j.gaitpost.2005.01.003

Hamner, S. R., Seth, A., & Delp, S. L. (2010). Muscle contributions to propulsion and support during running. *Journal of Biomechanics*, *43*(14), 2709–2716. https://doi.org/10.1016/j.jbiomech.2010.06.025

Harridge, S. D., Kryger, A., & Stensgaard, A. (1999). Knee extensor strength, activation, and size in very elderly people following strength training. *Muscle & Nerve*, *22*(7), 831–839. https://doi.org/10.1002/(sici)1097-4598(199907)22:7%253C831::aid-mus4%253E3.0.co;2-3

Hartholt, K. A., Van Beeck, E. F., Polinder, S., Van Der Velde, N., Van Lieshout, E. M. M., Panneman, M. J. M., Van Der Cammen, T. J. M., & Patka, P. (2011). Societal Consequences of Falls in the Older Population: Injuries, Healthcare Costs, and Long-Term Reduced Quality of Life. *Journal of Trauma: Injury, Infection & Critical Care*, *71*(3), 748–753. https://doi.org/10.1097/TA.0b013e3181f6f5e5

Hassan, A. S., Fajardo, M. E., Cummings, M., McPherson, L. M., Negro, F., Dewald, J. P. A., Heckman, C. J., & Pearcey, G. E. P. (2021a). Estimates of persistent inward currents are reduced in upper limb motor units of older adults. *Journal of Physiology*, *599*(21), 4865–4882. Scopus. https://doi.org/10.1113/JP282063

Hassan, A. S., Fajardo, M. E., Cummings, M., McPherson, L. M., Negro, F., Dewald, J. P. A., Heckman, C. J., & Pearcey, G. E. P. (2021b). Estimates of persistent inward currents are reduced in upper limb motor units of older adults. *Journal of Physiology*, *599*(21), 4865–4882. https://doi.org/10.1113/JP282063.

Haug, H., & Eggers, R. (1991). Morphometry of the human cortex cerebri and corpus striatum during aging. *Neurobiology of Aging*, *12*(4), 336–338. https://doi.org/10.1016/0197-4580(91)90013-A

Hausdorff, J. M., Zemany, L., Peng, C.-K., & Goldberger, A. L. (1999). Maturation of gait dynamics: Stride-to-stride variability and its temporal organization in children. *Journal of Applied Physiology*, *86*(3), 1040–1047. https://doi.org/10.1152/jappl.1999.86.3.1040

Haywood, K. M., & Getchell, N. (2021). *Life Span Motor Development* (6th Edition). Human Kinetics.

Heglund, N. C. (1981). A Simple Design for A Force-Plate to Measure Ground Reaction Forces. *Journal of Experimental Biology*, *93*(1), 333–338. https://doi.org/10.1242/jeb.93.1.333

Henderson, G., Tomlinson, B. E., & Gibson, P. H. (1980). Cell counts in human cerebral cortex in normal adults throughout life using an image analysing computer. *Journal of the Neurological Sciences*, *46*(1), 113–136. https://doi.org/10.1016/0022-510X(80)90048-9

Hepple, R. T., & Rice, C. L. (2016). Innervation and neuromuscular control in ageing skeletal muscle. *The Journal of Physiology*, *594*(8), 1965–1978. https://doi.org/10.1113/JP270561

Herman, T., Giladi, N., Gurevich, T., & Hausdorff, J. M. (2005). Gait instability and fractal dynamics of older adults with a "cautious" gait: Why do certain older adults walk fearfully? *Gait & Posture*, *21*(2), 178–185. https://doi.org/10.1016/j.gaitpost.2004.01.014

Hiebert, G. W., Whelan, P. J., Prochazka, A., & Pearson, K. G. (1996). Contribution of hind limb flexor muscle afferents to the timing of phase transitions in the cat step cycle. *Journal of Neurophysiology*, *75*(3), 1126–1137. https://doi.org/10.1152/jn.1996.75.3.1126

Hobo, K., Kawase, J., Tamura, F., Groher, M., Kikutani, T., & Sunakawa, H. (2014). Effects of the reappearance of primitive reflexes on eating function and prognosis. *Geriatrics & Gerontology International*, *14*(1), 190–197. https://doi.org/10.1111/ggi.12078

Hollman, J. H., McDade, E. M., & Petersen, R. C. (2011). Normative spatiotemporal gait parameters in older adults. *Gait & Posture*, *34*(1), 111–118. https://doi.org/10.1016/j.gaitpost.2011.03.024

Honda, K., Sekiguchi, Y., & Izumi, S.-I. (2023). Effect of Aging on the Trunk and Lower Limb Kinematics during Gait on a Compliant Surface in Healthy Individuals. *Biomechanics*, *3*(1), 103–114. https://doi.org/10.3390/biomechanics3010010

Honeine, J.-L., Schieppati, M., Gagey, O., & Do, M.-C. (2013). The Functional Role of the Triceps Surae Muscle during Human Locomotion. *PLoS ONE*, *8*(1), e52943. https://doi.org/10.1371/journal.pone.0052943

Hora, M., Struška, M., Matějovská, Z., Kubový, P., & Sládek, V. (2024). Muscle activity during crouched walking. *American Journal of Biological Anthropology*, *183*(1), 79–91. https://doi.org/10.1002/ajpa.24834

Horak, F. B. (2006). Postural orientation and equilibrium: What do we need to know about neural control of balance to prevent falls? *Age and Ageing*, *35 Suppl 2*, ii7–ii11. https://doi.org/10.1093/ageing/afl077

Hortobágyi, T., A, F., S, S., P, R., & P, D. (2011). Association between muscle activation and metabolic cost of walking in young and old adults. *The Journals of Gerontology. Series A, Biological Sciences and Medical Sciences*, *66*(5). https://doi.org/10.1093/gerona/glr008

Hortobágyi, T., & DeVita, P. (2006). Mechanisms responsible for the age-associated increase in coactivation of antagonist muscles. *Exercise and Sport Sciences Reviews*, *34*(1), 29–35. Scopus. https://doi.org/10.1097/00003677-200601000-00007

Hortobagyi, T., Finch, A., Solnik, S., Rider, P., & DeVita, P. (2011). Association Between Muscle Activation and Metabolic Cost of Walking in Young and Old Adults. *The Journals of Gerontology Series A: Biological Sciences and Medical Sciences*, *66A*(5), 541–547. https://doi.org/10.1093/gerona/glr008

Hortobágyi, T., Lesinski, M., Gäbler, M., VanSwearingen, J. M., Malatesta, D., & Granacher, U. (2015). Effects of Three Types of Exercise Interventions on Healthy Old Adults' Gait Speed: A Systematic Review and Meta-Analysis. *Sports Medicine*, *45*(12), 1627–1643. https://doi.org/10.1007/s40279-015-0371-2

Hortobágyi, T., Solnik, S., Gruber, A., Rider, P., Steinweg, K., Helseth, J., & DeVita, P. (2009). Interaction between age and gait velocity in the amplitude and timing of antagonist muscle coactivation. *Gait & Posture*, *29*(4), 558–564. https://doi.org/10.1016/j.gaitpost.2008.12.007

Hubel, T. Y., & Usherwood, J. R. (2015). Children and adults minimise activated muscle volume by selecting gait parameters that balance gross mechanical power and work demands. *Journal of Experimental Biology*, *218*(18), 2830–2839. https://doi.org/10.1242/jeb.122135

Hubli, M., Dietz, V., Schrafl-Altermatt, M., & Bolliger, M. (2013). Modulation of spinal neuronal excitability by spinal direct currents and locomotion after spinal cord injury. *Clinical Neurophysiology*, *124*(6), 1187–1195. https://doi.org/10.1016/j.clinph.2012.11.021

Hunter, G. R., Wetzstein, C. J., Fields, D. A., Brown, A., & Bamman, M. M. (2000). Resistance training increases total energy expenditure and free-living physical activity in older adults. *Journal of Applied Physiology*, *89*(3), 977–984. Scopus. https://doi.org/10.1152/jappl.2000.89.3.977

Hunter, L. C., Hendrix, E. C., & Dean, J. C. (2010). The cost of walking downhill: Is the preferred gait energetically optimal? *Journal of Biomechanics*, *43*(10), 1910–1915. https://doi.org/10.1016/j.jbiomech.2010.03.030

Hunter, S. K., Pereira, H. M., & Keenan, K. G. (2016). The aging neuromuscular system and motor performance. *Journal of Applied Physiology (Bethesda, Md.: 1985)*, *121*(4), 982–995. https://doi.org/10.1152/japplphysiol.00475.2016

Hyatt, R. H., Whitelaw, M. N., Bhat, A., Scott, S., & Maxwell, J. D. (1990). Association of muscle strength with functional status of elderly people. *Age and Ageing*, *19*(5), 330–336. Scopus. https://doi.org/10.1093/ageing/19.5.330

Ievins, A., & Moritz, C. T. (2017). Therapeutic Stimulation for Restoration of Function After Spinal Cord Injury. *Physiology*, *32*(5), 391–398. https://doi.org/10.1152/physiol.00010.2017

Ivanenko, Y. P., Cappellini, G., Poppele, R. E., & Lacquaniti, F. (2008). Spatiotemporal organization of α-motoneuron activity in the human spinal cord during different gaits and gait transitions. *European Journal of Neuroscience*, *27*(12), 3351–3368. https://doi.org/10.1111/j.1460-9568.2008.06289.x

Ivanenko, Y. P., Dominici, N., Cappellini, G., Dan, B., Cheron, G., & Lacquaniti, F. (2004). Development of pendulum mechanism and kinematic coordination from the first unsupported steps in toddlers. *Journal of Experimental Biology*, *207*(21), 3797–3810. https://doi.org/10.1242/jeb.01214

Ivanenko, Y. P., Dominici, N., Cappellini, G., Di Paolo, A., Giannini, C., Poppele, R. E., & Lacquaniti, F. (2013). Changes in the Spinal Segmental Motor Output for Stepping during Development from Infant to Adult. *The Journal of Neuroscience*, *33*(7), 3025–3036. https://doi.org/10.1523/JNEUROSCI.2722-12.2013

Ivanenko, Y. P., Dominici, N., Cappellini, G., & Lacquaniti, F. (2005). Kinematics in Newly Walking Toddlers Does Not Depend Upon Postural Stability. *Journal of Neurophysiology*, *94*(1), 754–763. https://doi.org/10.1152/jn.00088.2005

Ivanenko, Y. P., Dominici, N., & Lacquaniti, F. (2007). Development of Independent Walking in Toddlers. *Exercise and Sport Sciences Reviews*, *35*(2), 67–73. https://doi.org/10.1249/JES.0b013e31803eafa8

Ivanenko, Y. P., Poppele, R. E., & Lacquaniti, F. (2004). Five basic muscle activation patterns account for muscle activity during human locomotion: Basic muscle activation patterns. *The Journal of Physiology*, *556*(1), 267–282. https://doi.org/10.1113/jphysiol.2003.057174

Ivanenko, Y. P., Poppele, R. E., & Lacquaniti, F. (2006). Spinal Cord Maps of Spatiotemporal Alpha-Motoneuron Activation in Humans Walking at Different Speeds. *Journal of Neurophysiology*, *95*(2), 602–618. https://doi.org/10.1152/jn.00767.2005

Iwasaki, S., & Yamasoba, T. (2015). Dizziness and Imbalance in the Elderly: Age-related Decline in the Vestibular System. *Aging and Disease*, *6*(1), 38. https://doi.org/10.14336/AD.2014.0128

Jang, Y. C., & Van Remmen, H. (2011). Age-associated alterations of the neuromuscular junction. *Experimental Gerontology*, *46*(2–3), 193–198. https://doi.org/10.1016/j.exger.2010.08.029

Jankowska, E. (2017). Spinal control of motor outputs by intrinsic and externally induced electric field potentials. *Journal of Neurophysiology*, *118*(2), 1221–1234. https://doi.org/10.1152/jn.00169.2017

Johnson, R. T., O'Neill, M. C., & Umberger, B. R. (2022). The effects of posture on the three-dimensional gait mechanics of human walking in comparison with walking in bipedal chimpanzees. *Journal of Experimental Biology*, *225*(5), jeb243272. https://doi.org/10.1242/jeb.243272

Jones, E. J., Chiou, S., Atherton, P. J., Phillips, B. E., & Piasecki, M. (2022). Ageing and exercise-induced motor unit remodelling. *The Journal of Physiology*, *600*(8), 1839–1849. https://doi.org/10.1113/JP281726

Kadhiresan, V. A., Hassett, C. A., & Faulkner, J. A. (1996). Properties of single motor units in medial gastrocnemius muscles of adult and old rats. *The Journal of Physiology*, *493*(2), 543–552. https://doi.org/10.1113/jphysiol.1996.sp021402

Karamanidis, K., & Arampatzis, A. (2005). Mechanical and morphological properties of different muscle-tendon units in the lower extremity and running mechanics: Effect of aging and physical activity. *The Journal of Experimental Biology*, *208*(Pt 20), 3907–3923. https://doi.org/10.1242/jeb.01830

Karrer, T. M., Josef, A. K., Mata, R., Morris, E. D., & Samanez-Larkin, G. R. (2017). Reduced dopamine receptors and transporters but not synthesis capacity in normal aging adults: A meta-analysis. *Neurobiology of Aging*, *57*, 36–46. https://doi.org/10.1016/j.neurobiolaging.2017.05.006

Keating, C. J., Cabrera-Linares, J. C., Párraga-Montilla, J. A., Latorre-Román, P. A., Del Castillo, R. M., & García-Pinillos, F. (2021). Influence of Resistance Training on Gait & Balance Parameters in Older Adults: A Systematic Review. *International Journal of Environmental Research and Public Health*, *18*(4), 1759. https://doi.org/10.3390/ijerph18041759

Kelley, G. A., & Kelley, K. S. (2017). Is sarcopenia associated with an increased risk of all-cause mortality and functional disability? *Experimental Gerontology*, *96*, 100–103. https://doi.org/10.1016/j.exger.2017.06.008

Kendall, F., McCreary, E., Provance, P., Rodgers, M., & Romani, W. (2005). *Muscles. Testing and Function with Posture and Pain*. Lippincott Williams and Wilkins.

Kendall, F. P., McCreary Kendall, E., Provance Geise, P., Rodgers, M., & Romani, W. (2005). *Muscles: Testing and Function, with Posture and Pain* (5th ed.).

Kerrigan, D. C., Riley, P. O., Rogan, S., & Burke, D. T. (2000). Compensatory advantages of toe walking. *Archives of Physical Medicine and Rehabilitation*, *81*(1), 38–44.

Kiehn, O. (2006). Locomotor circuits in the mammalian spinal cord. *Annual Review of Neuroscience*, *29*, 279–306. https://doi.org/10.1146/%2520annurev.neuro.29.051605.112910

Klass, M., Baudry, S., & Duchateau, J. (2008). Age-related decline in rate of torque development is accompanied by lower maximal motor unit discharge frequency during fast contractions. *Journal of Applied Physiology*, *104*(3), 739–746. https://doi.org/10.1152/japplphysiol.00550.2007

Kluger, D., Major, M. J., Fatone, S., & Gard, S. A. (2014). The effect of trunk flexion on lower-limb kinetics of able-bodied gait. *Human*

Movement Science, *33*, 395–403. https://doi.org/10.1016/j.humov.2013.12.006

Koller, W. C. (1984). Primitive Reflexes in the Evaluation of the Aging Patient. *Clinical Gerontologist*, *3*(2), 19–22. https://doi.org/10.1300/J018v03n02_04

Kraemer, W. J., Adams, K., Cafarelli, E., Dudley, G. A., Dooly, C., Feigenbaum, M. S., Fleck, S. J., Franklin, B., Fry, A. C., Hoffman, J. R., Newton, R. U., Potteiger, J., Stone, M. H., Ratamess, N. A., & Triplett-McBride, T. (2002). Progression models in resistance training for healthy adults. *Medicine and Science in Sports and Exercise*, *34*(2), 364–380. https://doi.org/10.1097/00005768-200202000-00027

Kraemer, W. J., Ratamess, N. A., & French, D. N. (2002). Resistance Training for Health and Performance: *Current Sports Medicine Reports*, *1*(3), 165–171. https://doi.org/10.1249/00149619-200206000-00007

Kraus, W. E., Janz, K. F., Powell, K. E., Campbell, W. W., Jakicic, J. M., Troiano, R. P., Sprow, K., Torres, A., Piercy, K. L., & 2018 PHYSICAL ACTIVITY GUIDELINES ADVISORY COMMITTEE*. (2019). Daily Step Counts for Measuring Physical

Activity Exposure and Its Relation to Health. *Medicine and Science in Sports and Exercise*, *51*(6), 1206–1212. https://doi.org/10.1249/MSS.0000000000001932

Kuo, A. D., Donelan, J. M., & Ruina, A. (2005). Energetic Consequences of Walking Like an Inverted Pendulum: Step-to-Step Transitions: *Exercise and Sport Sciences Reviews*, *33*(2), 88–97. https://doi.org/10.1097/00003677-200504000-00006

Kwon, Y. N., & Yoon, S. S. (2017). Sarcopenia: Neurological Point of View. *Journal of Bone Metabolism*, *24*(2), 83. https://doi.org/10.11005/jbm.2017.24.2.83

La Scaleia, V., Ivanenko, Y. P., Zelik, K. E., & Lacquaniti, F. (2014). Spinal motor outputs during step-to-step transitions of diverse human gaits. *Frontiers in Human Neuroscience*, *8*. https://doi.org/10.3389/fnhum.2014.00305

Lacquaniti, F., Ivanenko, Y. P., & Zago, M. (2012). Patterned control of human locomotion: Control of human locomotion. *The Journal of Physiology*, *590*(10), 2189–2199. https://doi.org/10.1113/jphysiol.2011.215137

Landi, F., Liperoti, R., Russo, A., Giovannini, S., Tosato, M., Capoluongo, E., Bernabei, R., & Onder, G. (2012). Sarcopenia as a risk factor for falls in elderly individuals: Results from the ilSIRENTE study. *Clinical Nutrition*, *31*(5), 652–658. https://doi.org/10.1016/j.clnu.2012.02.007

Larsson, L., Sjödin, B., & Karlsson, J. (1978). Histochemical and biochemical changes in human skeletal muscle with age in sedentary males, age 22–65 years. *Acta Physiologica Scandinavica*, *103*(1), 31–39. https://doi.org/10.1111/j.1748-1716.1978.tb06187.x

Larsson, L., Xiaopeng, L., & Hans E., B. (1995). Effects of removal of weight.bearing function on contractility and myosin isoform composition in single human skeletal muscle cells. *European Journal of Physiology*, *432*, 320–328.

Lay, A. N., Hass, C. J., & Gregor, R. J. (2006). The effects of sloped surfaces on locomotion: A kinematic and kinetic analysis. *Journal of Biomechanics*, *39*(9), 1621–1628. https://doi.org/10.1016/j.jbiomech.2005.05.005

Lazarus, N. R., & Harridge, S. D. R. (2018). The Inherent Human Aging Process and the Facilitating Role of Exercise. *Frontiers in Physiology*, *9*, 1135. https://doi.org/10.3389/fphys.2018.01135

Lecker, S. H., Jagoe, R. T., Gilbert, A., Gomes, M., Baracos, V., Bailey, J., Price, S. R., Mitch, W. E., & Goldberg, A. L. (2004). Multiple types of skeletal muscle atrophy involve a common program of changes in gene expression. *The FASEB Journal*, *18*(1), 39–51. https://doi.org/10.1096/fj.03-0610com

Lee, S. J., & Hidler, J. (2008). Biomechanics of overground vs. Treadmill walking in healthy individuals. *Journal of Applied Physiology (Bethesda, Md.: 1985)*, *104*(3), 747–755. https://doi.org/10.1152/japplphysiol.01380.2006

Lenoir, C., Jankovski, A., & Mouraux, A. (2018). Anodal Transcutaneous Spinal Direct Current Stimulation (tsDCS) Selectively Inhibits the Synaptic Efficacy of Nociceptive Transmission at Spinal Cord Level. *Neuroscience*, *393*, 150–163. https://doi.org/10.1016/j.neuroscience.2018.10.007

Leteneur, S., Gillet, C., Sadeghi, H., Allard, P., & Barbier, F. (2009). Effect of trunk inclination on lower limb joint and lumbar moments in able men during the stance phase of gait. *Clinical Biomechanics*, *24*(2), 190–195. https://doi.org/10.1016/j.clinbiomech.2008.10.005

Lettner, J., Królikowska, A., Ramadanov, N., Oleksy, Ł., Hakam, H. T., Becker, R., & Prill, R. (2024). Evaluating the Reliability of MyotonPro in Assessing Muscle Properties: A Systematic Review of

Diagnostic Test Accuracy. *Medicina*, *60*(6), 851. https://doi.org/10.3390/medicina60060851

Levine, R. V., & Norenzayan, A. (1999). The Pace of Life in 31 Countries. *Journal of Cross-Cultural Psychology*, *30*(2), 178–205. https://doi.org/10.1177/0022022199030002003

Lexell, J. (1995). Human Aging, Muscle Mass, and Fiber Type Composition. *The Journal of Gerontology*, *50A*, 11–16. https://doi.org/10.1093/gerona/50A.Special_Issue.11

Lexell, J., Henriksson-Larsén, K., Winblad, B., & Sjöström, M. (1983). Distribution of different fiber types in human skeletal muscles: Effects of aging studied in whole muscle cross sections. *Muscle & Nerve*, *6*(8), 588–595. https://doi.org/10.1002/mus.880060809

Lexell, J., & Taylor, C. C. (1991b). Variability in muscle fibre areas in whole human quadriceps muscle: Effects of increasing age. *Journal of Anatomy*, *174*, 239–249.

Li Y., R. H. C., Alexander, R. M., Gunther, M. M., & Wang, W. J. (1996, February 14). Characteristics of Ground Reaction Forces in Normal and Chimpanzee-Like Bipedal Walking by Humans. *Folia Primatologica*, *66*(1–4), 137–159.

Lim, J.-Y., Choi, S. J., Widrick, J. J., Phillips, E. M., & Frontera, W. R. (2019). Passive force and viscoelastic properties of single fibers in human aging muscles. *European Journal of Applied Physiology*, *119*(10), 2339–2348. Scopus. https://doi.org/10.1007/s00421-019-04221-7

Liu, P., Hao, Q., Hai, S., Wang, H., Cao, L., & Dong, B. (2017). Sarcopenia as a predictor of all-cause mortality among community-dwelling older people: A systematic review and meta-analysis. *Maturitas*, *103*, 16–22. https://doi.org/10.1016/j.maturitas.2017.04.007

López-Otín, C., Blasco, M. A., Partridge, L., Serrano, M., & Kroemer, G. (2013). The Hallmarks of Aging. *Cell*, *153*(6), 1194–1217. https://doi.org/10.1016/j.cell.2013.05.039

Luc-Harkey, B. A., Franz, J. R., Blackburn, J. T., Padua, D. A., Hackney, A. C., & Pietrosimone, B. (2018). Real-time biofeedback can increase and decrease vertical ground reaction force, knee flexion excursion, and knee extension moment during walking in individuals with anterior cruciate ligament reconstruction. *Journal of Biomechanics*, *76*, 94–102. https://doi.org/10.1016/j.jbiomech.2018.05.043

Maier, E. (1961). Longitudinal measurement research on the maturation of the child's foot. *Monatsschr Kinderheilkd*, *109*, 222–226.

Malatesta, D., Simar, D., Saad, H. B., Préfaut, C., & Caillaud, C. (2010). Effect of an overground walking training on gait performance in healthy 65- to 80-year-olds. *Experimental Gerontology*, *45*(6), 427–434. Scopus. https://doi.org/10.1016/j.exger.2010.03.009

Mallapaty, S. (2024, December 12). What is Ageing? Even Scientists Can´t Agree: Survey of scientists who study ageing finds no consensus on the fundamentals. *Nature*, *636*, 283–284.

Malloggi, C., Rota, V., Catino, L., Malfitano, C., Scarano, S., Soranna, D., Zambon, A., & Tesio, L. (2019). Three-dimensional path of the body centre of mass during walking in children: An index of neural maturation. *International Journal of Rehabilitation Research*, *42*(2), 112–119. https://doi.org/10.1097/MRR.0000000000000345

Marcucci, L., & Reggiani, C. (2020). Increase of resting muscle stiffness, a less considered component of age-related skeletal muscle impairment. *European Journal of Translational Myology*, *30*(2), 8982. https://doi.org/10.4081/ejtm.2019.8982

Mari, S., Serrao, M., Casali, C., Conte, C., Martino, G., Ranavolo, A., Coppola, G., Draicchio, F., Padua, L., Sandrini, G., & Pierelli, F. (2014). Lower limb antagonist muscle co-activation and its relationship with gait parameters in cerebellar ataxia. *Cerebellum*

(London, England), *13*(2), 226–236. https://doi.org/10.1007/s12311-013-0533-4

Markov, A., Hauser, L., & Chaabene, H. (2022). Effects of Concurrent Strength and Endurance Training on Measures of Physical Fitness in Healthy Middle-Aged and Older Adults: A Systematic Review with Meta-Analysis. *Sports Medicine*, *53*(2), 437–455. https://doi.org/10.1007/s40279-022-01764-2

Marques, N. R., LaRoche, D. P., Hallal, C. Z., Crozara, L. F., Morcelli, M. H., Karuka, A. H., Navega, M. T., & Gonçalves, M. (2013). Association between energy cost of walking, muscle activation, and biomechanical parameters in older female fallers and non-fallers. *Clinical Biomechanics*, *28*(3), 330–336. https://doi.org/10.1016/j.clinbiomech.2013.01.004

Martino, G., Ivanenko, Y. P., d'Avella, A., Serrao, M., Ranavolo, A., Draicchio, F., Cappellini, G., Casali, C., & Lacquaniti, F. (2015). Neuromuscular adjustments of gait associated with unstable conditions. *Journal of Neurophysiology*, *114*(5), 2867–2882. https://doi.org/10.1152/jn.00029.2015

Martino, G., Ivanenko, Y. P., Serrao, M., Ranavolo, A., d'Avella, A., Draicchio, F., Conte, C., Casali, C., & Lacquaniti, F. (2014).

Locomotor patterns in cerebellar ataxia. *Journal of Neurophysiology*, *112*(11), 2810–2821. https://doi.org/10.1152/jn.00275.2014

Martino, G., Ivanenko, Y., Serrao, M., Ranavolo, A., Draicchio, F., Rinaldi, M., Casali, C., & Lacquaniti, F. (2018). Differential changes in the spinal segmental locomotor output in Hereditary Spastic Paraplegia. *Clinical Neurophysiology*, *129*(3), 516–525. https://doi.org/10.1016/j.clinph.2017.11.028

Massaad, F., Lejeune, T. M., & Detrembleur, C. (2007). The up and down bobbing of human walking: A compromise between muscle work and efficiency. *The Journal of Physiology*, *582*(2), 789–799. https://doi.org/10.1113/jphysiol.2007.127969

Mau-Moeller, A., Behrens, M., Lindner, T., Bader, R., & Bruhn, S. (2013). Age-related changes in neuromuscular function of the quadriceps muscle in physically active adults. *Journal of Electromyography and Kinesiology*, *23*(3), 640–648. https://doi.org/10.1016/j.jelekin.2013.01.009

McCollum, G., Holroyd, C., & Castelfranco M., A. (1995). Forms of Early Walking. *Journal of Theoretical Biology*, *176*(3), 373–390. https://doi.org/10.1006/jtbi.1995.0206

McGibbon, C. A. (2003). Toward a better understanding of gait changes with age and disablement: Neuromuscular adaptation. *Exercise and Sport Sciences Reviews*, *31*(2), 102–108. https://doi.org/10.1097/00003677-200304000-00009

McNeil, C. J., Doherty, T. J., Stashuk, D. W., & Rice, C. L. (2005). Motor unit number estimates in the tibialis anterior muscle of young, old, and very old men. *Muscle & Nerve*, *31*(4), 461–467. https://doi.org/10.1002/mus.20276

McPhee, J. S., Cameron, J., Maden-Wilkinson, T., Piasecki, M., Yap, M. H., Jones, D. A., & Degens, H. (2018). The Contributions of Fiber Atrophy, Fiber Loss, In Situ Specific Force, and Voluntary Activation to Weakness in Sarcopenia. *The Journals of Gerontology. Series A, Biological Sciences and Medical Sciences*, *73*(10), 1287–1294. https://doi.org/10.1093/gerona/gly040

McPhee, J. S., French, D. P., Jackson, D., Nazroo, J., Pendleton, N., & Degens, H. (2016). Physical activity in older age: Perspectives for healthy ageing and frailty. *Biogerontology*, *17*(3), 567–580. https://doi.org/10.1007/s10522-016-9641-0

Mesquita, R. M., Catavitello, G., Willems, P. A., & Dewolf, A. H. (2023). Modification of the locomotor pattern when deviating from the characteristic heel-to-toe rolling pattern during walking. *European Journal of Applied Physiology*, *123*(7), 1455–1467. https://doi.org/10.1007/s00421-023-05169-5

Mesquita, R. M., Dewolf, A. H., Catavitello, G., Osgnach, C., di Prampero, P. E., & Willems, P. A. (2020). The bouncing mechanism of running against hindering, or with aiding traction forces: A comparison with running on a slope. *European Journal of Applied Physiology*. https://doi.org/10.1007/s00421-020-04379-5

Metcalf, K. M., Baquero, B. I., Coronado Garcia, M. L., Francis, S. L., Janz, K. F., Laroche, H. H., & Sewell, D. K. (2018). Calibration of the global physical activity questionnaire to Accelerometry measured physical activity and sedentary behavior. *BMC Public Health*, *18*(1), 412. https://doi.org/10.1186/s12889-018-5310-3

Meurisse, G. M., Bastien, G. J., & Schepens, B. (2019a). Effect of age and speed on the step-to-step transition phase during walking. *Journal of Biomechanics*, *83*, 253–259. https://doi.org/10.1016/j.jbiomech.2018.12.001

Meurisse, G. M., Bastien, G. J., & Schepens, B. (2019b). The step-to-step transition mode: A potential indicator of first-fall risk in elderly adults? *PLOS ONE*, *14*(8), e0220791. https://doi.org/10.1371/journal.pone.0220791

Meurisse, G. M., Dierick, F., Schepens, B., & Bastien, G. J. (2016a). Determination of the vertical ground reaction forces acting upon individual limbs during healthy and clinical gait. *Gait & Posture*, *43*, 245–250. https://doi.org/10.1016/j.gaitpost.2015.10.005

Mian, O. S., Thom, J. M., Ardigò, L. P., Narici, M. V., & Minetti, A. E. (2006). Metabolic cost, mechanical work, and efficiency during walking in young and older men. *Acta Physiologica*, *186*(2), 127–139. https://doi.org/10.1111/j.1748-1716.2006.01522.x

Monaco, V., Ghionzoli, A., & Micera, S. (2010). Age-Related Modifications of Muscle Synergies and Spinal Cord Activity During Locomotion. *Journal of Neurophysiology*, *104*(4), 2092–2102. https://doi.org/10.1152/jn.00525.2009

Monaco, V., & Micera, S. (2012). Age-related neuromuscular adaptation does not affect the mechanical efficiency of lower limbs during walking. *Gait & Posture*, *36*(3), 350–355. https://doi.org/10.1016/j.gaitpost.2012.03.031

Montero-Odasso, M., Verghese, J., Beauchet, O., & Hausdorff, J. M. (2012). Gait and Cognition: A Complementary Approach to

Understanding Brain Function and the Risk of Falling. *Journal of the American Geriatrics Society*, *60*(11), 2127–2136. https://doi.org/10.1111/j.1532-5415.2012.04209.x

Mortaza, N., Abu Osman, N. A., & Mehdikhani, N. (2014). Are the spatio-temporal parameters of gait capable of distinguishing a faller from a non-faller elderly? *European Journal of Physical and Rehabilitation Medicine*, *50*(6), 677–691.

Mosole, S., Carraro, U., Kern, H., Loefler, S., Fruhmann, H., Vogelauer, M., Burggraf, S., Mayr, W., Krenn, M., Paternostro-Sluga, T., Hamar, D., Cvecka, J., Sedliak, M., Tirpakova, V., Sarabon, N., Musarò, A., Sandri, M., Protasi, F., Nori, A., … Zampieri, S. (2014). Long-Term High-Level Exercise Promotes Muscle Reinnervation With Age: *Journal of Neuropathology & Experimental Neurology*, *73*(4), 284–294. https://doi.org/10.1097/NEN.0000000000000032

Müller, R., Rode, C., Aminiaghdam, S., Vielemeyer, J., & Blickhan, R. (2017). Force direction patterns promote whole body stability even in hip-flexed walking, but not upper body stability in human upright walking. *Proceedings of the Royal Society A: Mathematical, Physical and Engineering Sciences*, *473*(2207), 20170404. https://doi.org/10.1098/rspa.2017.0404

Murton, A. J. (2015). Muscle protein turnover in the elderly and its potential contribution to the development of sarcopenia. *Proceedings of the Nutrition Society*, *74*(4), 387–396. https://doi.org/10.1017/S0029665115000130

Nguyen, A. P., Verdebout, J., Fontaine, A., Mahaudens, P., & Detrembleur, C. (2024). Recommendation for stiffness measurement on gastrocnemii and Achilles tendon at rest and in loaded conditions. *European Rehabilitation Journal*, *4*(1), Article 1. https://doi.org/10.52057/erj.v4i1.42

Noble, J. W., & Prentice, S. D. (2008). Intersegmental coordination while walking up inclined surfaces: Age and ramp angle effects. *Experimental Brain Research*, *189*(2), 249–255. https://doi.org/10.1007/s00221-008-1464-z

Nordin, F., Nyberg, A., & Sandberg, C. (2020). Concurrent validity of a fixated hand-held dynamometer for measuring isometric knee extension strength in adults with congenital heart disease. *European Journal of Physiotherapy*, *22*(4), 206–211. https://doi.org/10.1080/21679169.2019.1573920

Nunez-Lisboa, M., & Dewolf, A. H. (2023). Neuromechanical adjustments in human walking with bent posture. *Gait & Posture*, *105*, S32. https://doi.org/10.1016/j.gaitpost.2023.07.323

Oertel, G. (1986). Changes in human skeletal muscles due to ageing: Histological and histochemical observations on autopsy material. *Acta Neuropathologica*, *69*(3–4), 309–313. https://doi.org/10.1007/BF00688309

Orssatto, L. B. R., Borg, D. N., Blazevich, A. J., Sakugawa, R. L., Shield, A. J., & Trajano, G. S. (2021). Intrinsic motoneuron excitability is reduced in soleus and tibialis anterior of older adults. *GeroScience*, *43*(6), 2719–2735. https://doi.org/10.1007/s11357-021-00478-z

Orssatto, L. B. R., Borg, D. N., Pendrith, L., Blazevich, A. J., Shield, A. J., & Trajano, G. S. (2022). Do motoneuron discharge rates slow with aging? A systematic review and meta-analysis. *Mechanisms of Ageing and Development*, *203*. Scopus. https://doi.org/10.1016/j.mad.2022.111647

Orssatto, L. B. R., Mesquita, R. N. O., & Phillips, K. M. (2021). Looking at the bigger PICture: Understanding and counteracting the decline of persistent inward currents in older adults. *The Journal of Physiology*, *599*(23), 5137–5139. https://doi.org/10.1113/JP282370

Ortega, J. D., & Farley, C. T. (2007). Individual limb work does not explain the greater metabolic cost of walking in elderly adults. *Journal of Applied Physiology*, *102*(6), 2266–2273. Scopus. https://doi.org/10.1152/japplphysiol.00583.2006

Pardes, A., Beach, Z., Raja, H., Rodriguez, A., Freedman, B., & Soslowsky, L. (2017). Aging leads to inferior Achilles tendon mechanics and altered ankle function in rodents. *Journal of Biomechanics*, *60*, 30–38. https://doi.org/10.1016/j.jbiomech.2017.06.008

Payne, A. M., & Delbono, O. (2004). Neurogenesis of Excitation-Contraction Uncoupling in Aging Skeletal Muscle: *Exercise and Sport Sciences Reviews*, *32*(1), 36–40. https://doi.org/10.1097/00003677-200401000-00008

Payne, G. V., & Larry D., I. (2017). *Human Motor Development A Lifespan Approach* (9th Edition).

Pearson, S. J., Young, A., Macaluso, A., Devito, G., Nimmo, M. A., Cobbold, M., & Harridge, S. D. R. (2002). Muscle function in elite master weightlifters. *Medicine and Science in Sports and Exercise*, *34*(7), 1199–1206. Scopus. https://doi.org/10.1097/00005768-200207000-00023

Peterson, D. S., & Martin, P. E. (2010b). Effects of age and walking speed on coactivation and cost of walking in healthy adults. *Gait & Posture*, *31*(3), 355–359. https://doi.org/10.1016/j.gaitpost.2009.12.005

Peyré-Tartaruga, L. A., Dewolf, A. H., Di Prampero, P. E., Fábrica, G., Malatesta, D., Minetti, A. E., Monte, A., Pavei, G., Silva-Pereyra, V., Willems, P. A., & Zamparo, P. (2021). Mechanical work as a (key) determinant of energy cost in human locomotion: Recent findings and future directions. *Experimental Physiology*, *106*(9), 1897–1908. https://doi.org/10.1113/EP089313

Piasecki, M., Ireland, A., Coulson, J., Stashuk, D. W., Hamilton-Wright, A., Swiecicka, A., Rutter, M. K., McPhee, J. S., & Jones, D. A. (2016). Motor unit number estimates and neuromuscular transmission in the tibialis anterior of master athletes: Evidence that athletic older people are not spared from age-related motor unit remodeling. *Physiological Reports*, *4*(19), e12987. https://doi.org/10.14814/phy2.12987

Piasecki, M., Ireland, A., Piasecki, J., Stashuk, D. W., McPhee, J. S., & Jones, D. A. (2018). The reliability of methods to estimate the number and size of human motor units and their use with large limb muscles. *European Journal of Applied Physiology*, *118*(4), 767–775. https://doi.org/10.1007/s00421-018-3811-5

Pin, S., & Spini, D. (2016). Impact of falling on social participation and social support trajectories in a middle-aged and elderly European sample. *SSM - Population Health*, *2*, 382–389. https://doi.org/10.1016/j.ssmph.2016.05.004

Pongmala, C., Fabbri, M., Zibetti, M., Pitakpatapee, Y., Wangthumrong, T., Sangpeamsook, T., Srikajon, J., Srivanitchapoom, P., Youn, J., Cho, J. W., Kim, M., Zamil Shinawi, H. M., Obaid, M. T., Baumann, A., Margraf, N. G., Pona-Ferreira, F., Leitão, M., Lobo, T., Ferreira, J. J., … Artusi, C. A. (2022). Gait and axial postural abnormalities correlations in Parkinson's disease: A multicenter quantitative study. *Parkinsonism & Related Disorders*, *105*, 19–23. https://doi.org/10.1016/j.parkreldis.2022.10.026

Pour, M.-A. B., Joukar, S., Hovanloo, F., & Najafipour, H. (2017). Long-term Low-Intensity Endurance Exercise along with Blood-Flow Restriction Improves Muscle Mass and Neuromuscular Junction Compartments in Old Rats. *Iranian Journal of Medical Sciences*, *42*(6), 569–576.

Power, G. A., Dalton, B. H., Behm, D. G., Vandervoort, A. A., Doherty, T. J., & Rice, C. L. (2010). Motor Unit Number Estimates in Masters Runners: Use It or Lose It? *Medicine & Science in Sports & Exercise*, *42*(9), 1644–1650. https://doi.org/10.1249/MSS.0b013e3181d6f9e9

Preece, S. J., & Alghamdi, W. (2021). Increased trunk flexion may underlie elevated knee flexor activity in people with knee osteoarthritis. *The Knee*, *33*, 216–225. https://doi.org/10.1016/j.knee.2021.10.006

Price, C., Morrison, S. C., Hashmi, F., Phethean, J., & Nester, C. (2018). Biomechanics of the infant foot during the transition to independent walking: A narrative review. *Gait & Posture*, *59*, 140–146. https://doi.org/10.1016/j.gaitpost.2017.09.005

Prince, F., Corriveau, H., Hébert, R., & Winter, D. A. (1997). Gait in the elderly. *Gait & Posture*, *5*(2), 128–135. https://doi.org/10.1016/S0966-6362(97)01118-1

Radaelli, R., Trajano, G. S., Freitas, S. R., Izquierdo, M., Cadore, E. L., & Pinto, R. S. (2023). Power Training Prescription in Older Individuals: Is It Safe and Effective to Promote Neuromuscular Functional Improvements? *Sports Medicine*, *53*(3), 569–576. https://doi.org/10.1007/s40279-022-01758-0

Raffalt, P. C., Alkjær, T., & Simonsen, E. B. (2015). Changes in soleus H-reflex during walking in middle-aged, healthy subjects. *Muscle & Nerve*, *51*(3), 419–425. https://doi.org/10.1002/mus.24279

Ransmayr, G., Faucheux, B., Nowakowski, C., Kubis, N., Federspiel, S., Kaufmann, W., Henin, D., Hauw, J.-J., Agid, Y., & Hirsch, E. C. (2000). Age-related changes of neuronal counts in the human pedunculopontine nucleus. *Neuroscience Letters*, *288*(3), 195–198. https://doi.org/10.1016/S0304-3940(00)01244-1

Reid, K. F., & Fielding, R. A. (2012). Skeletal Muscle Power: A Critical Determinant of Physical Functioning In Older Adults. *Exercise and Sport Sciences Reviews*, *40*(1), 4–12. https://doi.org/10.1097/JES.0b013e31823b5f13

Robinovitch, S. N., Feldman, F., Yang, Y., Schonnop, R., Leung, P. M., Sarraf, T., Sims-Gould, J., & Loughin, M. (2013). Video capture of the circumstances of falls in elderly people residing in long-term care: An observational study. *The Lancet*, *381*(9860), 47–54. https://doi.org/10.1016/S0140-6736(12)61263-X

Rogers, R. S., & Nishimune, H. (2017). THE ROLE OF LAMININS IN THE ORGANIZATION AND FUNCTION OF NEUROMUSCULAR JUNCTIONS. *Matrix Biology : Journal of the*

International Society for Matrix Biology, *57–58*, 86–105. https://doi.org/10.1016/j.matbio.2016.08.008

Rudolph, K. S., Axe, M. J., & Snyder-Mackler, L. (2000). Dynamic stability after ACL injury: Who can hop? *Knee Surgery, Sports Traumatology, Arthroscopy: Official Journal of the ESSKA*, *8*(5), 262–269.

Ruina, A., Bertram, J. E. A., & Srinivasan, M. (2005). A collisional model of the energetic cost of support work qualitatively explains leg sequencing in walking and galloping, pseudo-elastic leg behavior in running and the walk-to-run transition. *Journal of Theoretical Biology*, *237*(2), 170–192. https://doi.org/10.1016/j.jtbi.2005.04.004

Rygiel, K. A., Picard, M., & Turnbull, D. M. (2016a). The ageing neuromuscular system and sarcopenia: A mitochondrial perspective. *The Journal of Physiology*, *594*(16), 4499–4512. https://doi.org/10.1113/JP271212

Scaglioni, G., Ferri, A., Minetti, A. E., Martin, A., Van Hoecke, J., Capodaglio, P., Sartorio, A., & Narici, M. V. (2002). Plantar flexor activation capacity and H reflex in older adults: Adaptations to strength training. *Journal of Applied Physiology*, *92*(6), 2292–2302. https://doi.org/10.1152/japplphysiol.00367.2001

Schepens, B., Bastien, G. J., Heglund, N. C., & Willems, P. A. (2004). Mechanical work and muscular efficiency in walking children. *Journal of Experimental Biology*, *207*(4), 587–596. https://doi.org/10.1242/jeb.00793

Schepens, B., Willems, P. A., Cavagna, G. A., & Heglund, N. C. (2001). Mechanical power and efficiency in running children. *Pflügers Archiv*, *442*(1), 107–116. https://doi.org/10.1007/s004240000511

Schmitz, A., Silder, A., Heiderscheit, B., Mahoney, J., & Thelen, D. G. (2009). Differences in lower-extremity muscular activation during walking between healthy older and young adults. *Journal of Electromyography and Kinesiology: Official Journal of the International Society of Electrophysiological Kinesiology*, *19*(6), 1085–1091. https://doi.org/10.1016/j.jelekin.2008.10.008

Şendur, H. N., Cindil, E., Cerit, M. N., Kılıç, P., Gültekin, I. İ., & Oktar, S. Ö. (2020). Evaluation of effects of aging on skeletal muscle elasticity using shear wave elastography. *European Journal of Radiology*, *128*, 109038. https://doi.org/10.1016/j.ejrad.2020.109038

Shefer, G., Van De Mark, D. P., Richardson, J. B., & Yablonka-Reuveni, Z. (2006). Satellite-cell pool size does matter: Defining the myogenic potency of aging skeletal muscle. *Developmental Biology*, *294*(1), 50–66. https://doi.org/10.1016/j.ydbio.2006.02.022

Sheth, K. A., Iyer, C. C., Wier, C. G., Crum, A. E., Bratasz, A., Kolb, S. J., Clark, B. C., Burghes, A. H. M., & Arnold, W. D. (2018). Muscle strength and size are associated with motor unit connectivity in aged mice. *Neurobiology of Aging*, *67*, 128–136. https://doi.org/10.1016/j.neurobiolaging.2018.03.016

Shigemoto, K., Kubo, S., Mori, S., Yamada, S., Akiyoshi, T., & Miyazaki, T. (2010). Muscle weakness and neuromuscular junctions in aging and disease. *Geriatrics & Gerontology International*, *10*(s1). https://doi.org/10.1111/j.1447-0594.2010.00608.x

Silder, A., Heiderscheit, B., & Thelen, D. G. (2008). Active and passive contributions to joint kinetics during walking in older adults. *Journal of Biomechanics*, *41*(7), 1520–1527. Scopus. https://doi.org/10.1016/j.jbiomech.2008.02.016

Siparsky, P. N., Kirkendall, D. T., & Garrett, W. E. (2014). Muscle Changes in Aging: Understanding Sarcopenia. *Sports Health: A Multidisciplinary Approach*, *6*(1), 36–40. https://doi.org/10.1177/1941738113502296

Skiadopoulos, A., & Knikou, M. (2024). Tapping into the human spinal locomotor centres with transspinal stimulation. *Scientific Reports*, *14*(1), 5990. https://doi.org/10.1038/s41598-024-56579-0

Sloot, L. H., Malheiros, S., Truijen, S., Saeys, W., Mombaur, K., Hallemans, A., & Van Criekinge, T. (2021). Decline in gait propulsion in older adults over age decades. *Gait & Posture*, *90*, 475–482. https://doi.org/10.1016/j.gaitpost.2021.09.166

Solopova, I. A., Sukhotina, I. A., Zhvansky, D. S., Ikoeva, G. A., Vissarionov, S. V., Baindurashvili, A. G., Edgerton, V. R., Gerasimenko, Y. P., & Moshonkina, T. R. (2017). Effects of spinal cord stimulation on motor functions in children with cerebral palsy. *Neuroscience Letters*, *639*, 192–198. https://doi.org/10.1016/j.neulet.2017.01.003

Song, M.-Y., Ruts, E., Kim, J., Janumala, I., Heymsfield, S., & Gallagher, D. (2004). Sarcopenia and increased adipose tissue infiltration of muscle in elderly African American women. *The American Journal of Clinical Nutrition*, *79*(5), 874–880. https://doi.org/10.1093/ajcn/79.5.874

Song, S., & Geyer, H. (2018). Predictive neuromechanical simulations indicate why walking performance declines with ageing. *Journal of*

Physiology, *596*(7), 1199–1210. Scopus. https://doi.org/10.1113/JP275166

Song, W., & Martin, J. H. (2017). Spinal cord direct current stimulation differentially modulates neuronal activity in the dorsal and ventral spinal cord. *Journal of Neurophysiology*, *117*(3), 1143–1155. https://doi.org/10.1152/jn.00584.2016

Spoor, F., Wod, B., & Zonneveld, F. (1994, May 25). Implications of early hominid labyrinthine morphology for evolution of human bipedal locomotion. *Letters to Nature*, *369*, 645–648.

Stalberg, E., & Fawcett, P. R. (1982). Macro EMG in healthy subjects of different ages. *Journal of Neurology, Neurosurgery & Psychiatry*, *45*(10), 870–878. https://doi.org/10.1136/jnnp.45.10.870

Stenroth, L., Peltonen, J., Cronin, N. J., Sipilä, S., & Finni, T. (2012). Age-related differences in Achilles tendon properties and triceps surae muscle architecture in vivo. *Journal of Applied Physiology*, *113*(10), 1537–1544. https://doi.org/10.1152/japplphysiol.00782.2012

Stock, M. S., Thompson, B. J., & Sobolewski, E. J. (2019). Age-Related Differences in the Predictability of Fast Gait Speed with Absolute and Rapid Squat Strength. *Journal of Science in Sport and Exercise*, *1*(3), 273–280. https://doi.org/10.1007/s42978-019-00032-6

Studenski, S. (2011). Gait Speed and Survival in Older Adults. *JAMA*, *305*(1), 50. https://doi.org/10.1001/jama.2010.1923

Sutherland, D. (1997). The development of mature gait. *Gait and Posture*.

Suzuki, M., Okamura, T., Shimazu, Y., Takahashi, H., Eguchi, K., Kano, K., & Tsuchiya, S. (1992). A study of falls experienced by institutionalized elderly. *Nihon Koshu Eisei Zasshi*, *39*(12), 927–940.

Sylos-Labini, F., La Scaleia, V., Cappellini, G., Dewolf, A., Fabiano, A., Solopova, I. A., Mondì, V., Ivanenko, Y., & Lacquaniti, F. (2022). Complexity of modular neuromuscular control increases and variability decreases during human locomotor development. *Communications Biology*, *5*(1), 1256. https://doi.org/10.1038/s42003-022-04225-8

Sylos-Labini, F., Magnani, S., Cappellini, G., La Scaleia, V., Fabiano, A., Picone, S., Paolillo, P., Di Paolo, A., Lacquaniti, F., & Ivanenko, Y. (2017). Foot Placement Characteristics and Plantar Pressure Distribution Patterns during Stepping on Ground in Neonates. *Frontiers in Physiology*, *8*, 784. https://doi.org/10.3389/fphys.2017.00784

Tajali, S., Balbinot, G., Pakosh, M., Sayenko, D. G., Zariffa, J., & Masani, K. (2024). Modulations in neural pathways excitability post transcutaneous spinal cord stimulation among individuals with spinal cord injury: A systematic review. *Frontiers in Neuroscience*, *18*. https://doi.org/10.3389/fnins.2024.1372222

Takakusaki, K. (2017). Functional Neuroanatomy for Posture and Gait Control. *Journal of Movement Disorders*, *10*(1), 1–17. https://doi.org/10.14802/jmd.16062

Tangen, E. M., Gjestvang, C., Stensrud, T., & Haakstad, L. A. H. (2022). Is there an association between total physical activity level and VO2max among fitness club members? A cross-sectional study. *BMC Sports Science, Medicine and Rehabilitation*, *14*(1), 109. https://doi.org/10.1186/s13102-022-00503-4

Taylor, C., McHugh, C., Mockler, D., Minogue, C., Reilly, R. B., & Fleming, N. (2021). Transcutaneous spinal cord stimulation and motor responses in individuals with spinal cord injury: A methodological review. *PLOS ONE*, *16*(11), e0260166. https://doi.org/10.1371/journal.pone.0260166

Taylor, D. (2014). Physical activity is medicine for older adults. *Postgraduate Medical Journal*, *90*(1059), 26–32. https://doi.org/10.1136/postgradmedj-2012-131366

Thelen, E., & Fisher, D. M. (1995). Newborn Stepping: An Explanation for a"Disappearing" Reflex. *Development Psychology*, *18*(5), 760–775. https://doi.org/10.1037/0012-1649.18.5.760

Tremblay, M. S., Aubert, S., Barnes, J. D., Saunders, T. J., Carson, V., Latimer-Cheung, A. E., Chastin, S. F. M., Altenburg, T. M., & Chinapaw, M. J. M. (2017). Sedentary Behavior Research Network (SBRN) – Terminology Consensus Project process and outcome. *International Journal of Behavioral Nutrition and Physical Activity*, *14*(1). https://doi.org/10.1186/s12966-017-0525-8

Troiano, R. P., McClain, J. J., Brychta, R. J., & Chen, K. Y. (2014). Evolution of accelerometer methods for physical activity research. *British Journal of Sports Medicine*, *48*(13), 1019–1023. https://doi.org/10.1136/bjsports-2014-093546

Usherwood, J. R., Channon, A. J., Myatt, J. P., Rankin, J. W., & Hubel, T. Y. (2012). The human foot and heel-sole-toe walking strategy: A mechanism enabling an inverted pendular gait with low isometric muscle force? *Journal of the Royal Society Interface*, *9*(75), 2396–2402. Scopus. https://doi.org/10.1098/rsif.2012.0179

Vaishya, R., & Vaish, A. (2020). Falls in Older Adults are Serious. *Indian Journal of Orthopaedics*, *54*(1), 69–74. https://doi.org/10.1007/s43465-019-00037-x

Valenti, G., Bonomi, A. G., & Westerterp, K. R. (2016). Multicomponent Fitness Training Improves Walking Economy in Older Adults. *Medicine and Science in Sports and Exercise*, *48*(7), 1365–1370. Scopus. https://doi.org/10.1249/MSS.0000000000000893

Van Boxtel, M. P. J., Bosma, H., Jolles, J., & Vreeling, F. W. (2006). Prevalence of primitive reflexes and the relationship with cognitive change in healthy adults: A report from the Maastricht Aging Study. *Journal of Neurology*, *253*(7), 935–941. https://doi.org/10.1007/s00415-006-0138-7

Van Hooren, B., Fuller, J. T., Buckley, J. D., Miller, J. R., Sewell, K., Rao, G., Barton, C., Bishop, C., & Willy, R. W. (2020). Is Motorized Treadmill Running Biomechanically Comparable to Overground Running? A Systematic Review and Meta-Analysis of Cross-Over Studies. *Sports Medicine (Auckland, N.Z.)*, *50*(4), 785–813. https://doi.org/10.1007/s40279-019-01237-z

VanSwearingen, J. M., & Studenski, S. A. (2014). Aging, Motor Skill, and the Energy Cost of Walking: Implications for the Prevention and Treatment of Mobility Decline in Older Persons. *The Journals of Gerontology Series A: Biological Sciences and Medical Sciences*, *69*(11), 1429–1436. https://doi.org/10.1093/gerona/glu153

Venturelli, M., Cè, E., Limonta, E., Schena, F., Caimi, B., Carugo, S., Veicsteinas, A., & Esposito, F. (2015). Effects of endurance, circuit, and relaxing training on cardiovascular risk factors in hypertensive elderly patients. *Age*, *37*(5). Scopus. https://doi.org/10.1007/s11357-015-9835-4

Verghese, J., Holtzer, R., Lipton, R. B., & Wang, C. (2009). Quantitative Gait Markers and Incident Fall Risk in Older Adults. *The Journals of Gerontology Series A: Biological Sciences and Medical Sciences*, *64A*(8), 896–901. https://doi.org/10.1093/gerona/glp033

Visser, M., & Schaap, L. A. (2011). Consequences of Sarcopenia. *Clinics in Geriatric Medicine*, *27*(3), 387–399. https://doi.org/10.1016/j.cger.2011.03.006

Waanders, J. B., Hortobágyi, T., Murgia, A., Devita, P., & Franz, J. R. (2019). Advanced Age Redistributes Positive but Not Negative Leg Joint Work during Walking. *Medicine & Science in Sports &*

Exercise, *51*(4), 615–623. https://doi.org/10.1249/MSS.0000000000001828

Wagner, F. B., Mignardot, J.-B., Le Goff-Mignardot, C. G., Demesmaeker, R., Komi, S., Capogrosso, M., Rowald, A., Seáñez, I., Caban, M., Pirondini, E., Vat, M., McCracken, L. A., Heimgartner, R., Fodor, I., Watrin, A., Seguin, P., Paoles, E., Van Den Keybus, K., Eberle, G., … Courtine, G. (2018). Targeted neurotechnology restores walking in humans with spinal cord injury. *Nature*, *563*(7729), 65–71. https://doi.org/10.1038/s41586-018-0649-2

Wang, Q., Cui, C., Zhang, N., Lin, W., Chai, S., Chow, S. K.-H., Wong, R. M. Y., Hu, Y., Law, S. W., & Cheung, W.-H. (2024). Effects of physical exercise on neuromuscular junction degeneration during ageing: A systematic review. *Journal of Orthopaedic Translation*, *46*, 91–102. https://doi.org/10.1016/j.jot.2024.03.007

Wang, Y., Watanabe, K., & Asaka, T. (2019). Effect of dance on multi-muscle synergies in older adults: A cross-sectional study. *BMC Geriatrics*, *19*(1), 340. https://doi.org/10.1186/s12877-019-1365-y

Warrener, A., Tamai, R., & Lieberman, D. E. (2021). The effect of trunk flexion angle on lower limb mechanics during running. *Human*

Movement Science, *78*, 102817. https://doi.org/10.1016/j.humov.2021.102817

Wilkinson, D. J., Piasecki, M., & Atherton, P. J. (2018). The age-related loss of skeletal muscle mass and function: Measurement and physiology of muscle fibre atrophy and muscle fibre loss in humans. *Ageing Research Reviews*, *47*, 123–132. https://doi.org/10.1016/j.arr.2018.07.005

Willems, P. A., Cavagna, G. A., & Heglund, N. C. (1995). External, Internal and Total Work in Human Locomotion. *Journal of Experimental Biology*, *198*(2), 379–393. https://doi.org/10.1242/jeb.198.2.379

Willems, P. A., & Gosseye, T. P. (2013). Does an instrumented treadmill correctly measure the ground reaction forces? *Biology Open*, *2*(12), 1421–1424. https://doi.org/10.1242/bio.20136379

Williamson, D. L., Godard, M. P., Porter, D. A., Costill, D. L., & Trappe, S. W. (2000). Progressive resistance training reduces myosin heavy chain coexpression in single muscle fibers from older men. *Journal of Applied Physiology*, *88*(2), 627–633. Scopus. https://doi.org/10.1152/jappl.2000.88.2.627

Winter, D. A. (2009). *Biomechanics and motor control of human movement* (4th ed). Wiley.

Winter, D. A., Patia, A. E., Frank, J. S., & Walt, S. E. (1990, January 19). Biomechanical Walking Pattern Changes in the Fit and Healthy Elderly. *Physical Therapy*, *70*(6), 340–347.

Winter, D. A., Patla, A. E., Frank, J. S., & Walt, S. E. (1990). Biomechanical walking pattern changes in the fit and healthy elderly. *Physical Therapy*, *70*(6), 340–347. https://doi.org/10.1093/ptj/70.6.340

World Health Organization. (2015). *World report on Ageing and Health*.

Wright, S. P., & Wolfinger, R. D. (1997). Repeated Measures Analysis Using Mixed Models: Some Simulation Results. In T. G. Gregoire, D. R. Brillinger, P. J. Diggle, E. Russek-Cohen, W. G. Warren, & R. D. Wolfinger (Eds.), *Modelling Longitudinal and Spatially Correlated Data* (Vol. 122, pp. 147–157). Springer New York. https://doi.org/10.1007/978-1-4612-0699-6_13

Wu, J., Wang, X., Ye, M., Wang, L., & Zheng, G. (2021). Effect of regular resistance training on memory in older adults: A systematic review. *Experimental Gerontology*, *150*, 111396. https://doi.org/10.1016/j.exger.2021.111396

Yamaguchi, T., Beck, M. M., Therkildsen, E. R., Svane, C., Forman, C., Lorentzen, J., Conway, B. A., Lundbye-Jensen, J., Geertsen, S. S., &

Nielsen, J. B. (2020). Transcutaneous spinal direct current stimulation increases corticospinal transmission and enhances voluntary motor output in humans. *Physiological Reports*, *8*(16). https://doi.org/10.14814/phy2.14531

Yang, J. F., Stephens, M. J., & Vishram, R. (1998). Infant stepping: A method to study the sensory control of human walking. *The Journal of Physiology*, *507*(3), 927–937. https://doi.org/10.1111/j.1469-7793.1998.927bs.x

Yokoyama, H., Ogawa, T., Shinya, M., Kawashima, N., & Nakazawa, K. (2017). Speed dependency in α-motoneuron activity and locomotor modules in human locomotion: Indirect evidence for phylogenetically conserved spinal circuits. *Proceedings of the Royal Society B: Biological Sciences*, *284*(1851), 20170290. https://doi.org/10.1098/rspb.2017.0290

Yue, G. H., Ranganathan, V. K., Siemionow, V., Liu, J. Z., & Sahgal, V. (1999). Older Adults Exhibit a Reduced Ability to Fully Activate Their Biceps Brachii Muscle. *The Journals of Gerontology Series A: Biological Sciences and Medical Sciences*, *54*(5), M249–M253. https://doi.org/10.1093/gerona/54.5.M249

Zhvansky, D. S., Sylos-Labini, F., Dewolf, A., Cappellini, G., d'Avella, A., Lacquaniti, F., & Ivanenko, Y. (2022). Evaluation of Spatiotemporal Patterns of the Spinal Muscle Coordination Output during Walking in the Exoskeleton. *Sensors (Basel, Switzerland)*, *22*(15), 5708. https://doi.org/10.3390/s22155708

Supplementary Chapter I

Effect of age and speed on the step-to-step transition strategies in children

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List of Abbreviations

CoM:	Center of Mass
FC:	Foot Contact
TO:	Toe Off
TDC:	Double Contact Phase Duration
F_v:	Vertical Ground Reaction Force
F_{back}:	Peak Vertical Force of the Back Leg
F_{front}:	Peak Vertical Force of the Front Leg
V_v:	Vertical Velocity of the Center of Mass
$V_{v,min}$:	Minimum Vertical Velocity of the Center of Mass
Fr:	Froude Number
T:	Stride Period

Abstract

Background: The development and acquisition of mature walking in children is multifactorial, depending among others on foot interaction with the ground, body dynamics and the knowledge of the 'rules' stemming from the gravity field. Indeed, each step the velocity of the center of mass must be redirected upwards. This redirection may be initiated by the trailing leg, propulsing forward and upward the body before foot contact, or later by the loading limb after the contact with the ground. While it has been suggested that mature walking develops slowly from first independent steps to about 7 years of age, it is still unknown how children acquire the appropriate loading and propulsion forces during the step-to-step transition. To answer that question, twenty-four children (from 3 to 12 years old) and twelve young adults (from 20 to 27 years old) walked on force platforms at different walking speed. The ground reaction forces under each foot were recorded and the vertical velocity of the centre of mass of the body was computed. With decreasing age and increasing velocity (or Froude number), the occurrence of unanticipated transition is higher, related to a different ratio between the vertical support of the front and back leg. The different transition strategy observed in children indicates that body weight transfer from one limb to the other is not fully mature at 12 years old.

Introduction

Foot-support interactions and gravity have a profound influence on optimization and energetics of walking and its stability. When children start to walk independently, they need to learn the 'rules' of walking in the gravity field, in particular, the pendulum mechanism of walking (Dominici et al., 2007; Ivanenko, Dominici, et al., 2004; Ivanenko et al., 2007). The step-to-step transition strategies are also tightly related to the effect of gravity on walking since they require appropriate loading and propulsion forces at the end of stance. The simple integration of stepping movements with forward translation can be found in the typical strategy of gait initiation in toddlers, whereby gait is initiated by letting the body fall forward, and in a form of early walking, whereby the toddler puts the swing foot forward to brake the fall and bring the body back to the original stance foot (McCollum et al., 1995). McCollum et al identify different types of walking in toddlers corresponding to different forms of early walking, the "twister", the "faller" and the "stepper". Often, the "faller" cannot stop without something to bump into, such as a wall or friendly adult (McCollum et al., 1995). This suggests that toddlers do not anticipate this event and fail to redirect the centre of mass of the body (CoM) before double stance, as adults do. This is also supported by the fact that, in toddlers, virtually no power is generated at the ankle joint before foot-off (Forssberg, 1985; Hallemans et al., 2006). Older children seem to learn the adult-like mechanics and exchange between potential and kinematic energy of the CoM when walking in the gravity field (Cavagna et al., 1983), however, it is not clear to what extent and at what age they also adopt the adult-like gravity-related anticipatory step- to-step transition strategy.

During the acquisition of independent and mature gait, the physical dynamics of the mechanical system (inertia, viscoelastic properties of muscles and tendons and body size) may affect the locomotor behavior (Adolph et al., 2018; Adolph & Robinson, 2015; A. H. Dewolf et al., 2020; Thelen & Fisher, 1995). For example, body size and proportions change dramatically during growth (Haywood & Getchell, 2021), modifying the location of the center of gravity (G. V. Payne & Larry D., 2017), and in turn its stability. Also, walking in toddlers is characterized by high variability of movements, legs wide apart, short steps, and poor balance over the single support leg (Adolph & Robinson, 2015; Dominici et al., 2011). The body's control balance is assumed to be a milestone for mature walking, and it is not fully acquired before 6 years old (Brenière & Bril, 1998). Furthermore, both in terms of mechanical energy and kinematics, the ability to walk develops slowly from first independent steps to about 7 years of age (Cheron, Bengoetxea, et al., 2001; Cheron, Bouillot, et al., 2001; Dominici et al., 2010; Ivanenko, Dominici, et al., 2004; Ivanenko et al., 2005, 2007), reflecting the progressive, dynamic integration of postural equilibrium and forward propulsion in a gravity-centered frame with age. Changes in limb coordination, center of mass dynamics and balance control may in turn affect the coordinated transfer of weight from the trailing limb to the leading one during the step-to-step transition of walking, most likely not fully ac- quired before 7 years old.

Also, foot structure in infants differs from adults (Bosch et al., 2007; Maier, 1961; Price et al., 2018), with for example the presence of a fat pad underneath the foot plantar surface during the first years after birth (Bertsch et al., 2004; Gould et al., 1989). Another interesting observation is the modification of foot contact strategy with age (Hallemans et al., 2006). In toddlers walking, the foot interaction with the ground is still immature because touch-down is with flat-foot or toes-first (G. V. Payne & Larry D., 2017), instead of using the heel-to-toe rolling pattern as in adults. The latter foot contact strategy is progressively adopted (after around 23 months of independent walking (G. V. Payne & Larry D., 2017). This latter factor is especially important since the heel-to-toe rolling pattern has been related to specific muscle activation patterns and strategy of step-to-step transition (Mesquita et al., 2023).

During the transition from one limb to another, the downward trajectory of the CoM has to be redirected to an upward trajectory (Bastien et al., 2003; Kuo et al., 2005; Meurisse et al., 2019b) . In adult walking, the propulsion of the trailing limb redirects the trajectory of the CoM before the contact of the leading limb. This so-called anticipated transition reduces the impact of the leading limb with the ground. The aim of this study is to investigate the development of such transitions during gait in children as compared to healthy adults, expanding the analysis of the effect of age on the mechanics of walking. Data of children from 3 to 12 years old walking at speeds from 0.27 to 2.5 m s^{-1} were compared to adults. The forces exerted by each leg were measured and analyzed during the step–to–step transition. Since several gait parameters (e.g. mechanical work, step frequency and variability) exhibit changes in children up to 12 years (Bastien et al., 2003; Hausdorff et al., 1999; Schepens et al., 2004), we hypothesize that younger children would exert less vertical force on the trailing limb and more on the leading limb during the transition than older children and adults, as they progressively deviate from the form of early walking observed in toddlers. We also anticipate a progressive anticipated step-to-step transition with increasing age.

Materials and methods

Subjects and experimental procedures

Data were collected at the same time and on the same subjects as in the study of Schepens et al. (Schepens et al., 2001) and the steps analyzed are the same as in the study of Bastien et al., (2003)(Bastien et al., 2003). Experiments were performed on 24 healthy children. For the sake of clarity, in the figures the children were arbitrary divided in five age groups, defined as follows: 3–4, 5–6, 7–8, 9–10, and 11–12 years-old. The 3–4- year-old group included subjects 3·years to <5·years old; the 5–6-year- old group included subjects 5·years to <7·years old, the 7–8-year-old group included subjects 7·years to <9·years old, the 9–10-year-old group included subjects 9·years to <11·years old and the 11–12-year-old group included subjects 11·years to <13·years old. Twelve healthy young adults of 20–27·years of age were also registered. The mean characteristics of each age group are given in Table S.1. Informed written consent of the subjects and/or their parents was obtained. The experiments were performed according to the Declaration of Helsinki and were approved by the local ethics committee. All of the subjects wore swimming suits and gym shoes. Subjects were asked to walk across a force platform at different speeds. The individual signals of the force plates data were collected independently with two different set-ups: either sixteen 1000 mm × 1000 mm force plates as described in Genin et al. (2010), ten 600 mm × 400 mm plates, similar to those described by (Heglund, 1981). The plates were sensitive to forces in the fore-aft and vertical directions and had a linear response to within 1 % of the measured value for forces up to 3000·N. The difference in the electrical signal to a given force applied at different points on the surface of the plates was <1 %. The crosstalk between the vertical and forward axis was <1 % of the applied force. The individual signals of the plates were digitized by a 12-bit analogue-to-digital converter every 5·ms and processed by means of a desktop computer. The mean speed was measured by two photocells placed at the level of the neck and set 1.5–5.0·m apart depending upon the speed. In each age group, the data were gathered into speed classes of 0.28·m·s—1 (1·km·h—1). In most cases, two trials per subject were recorded in each speed class.

Data analysis

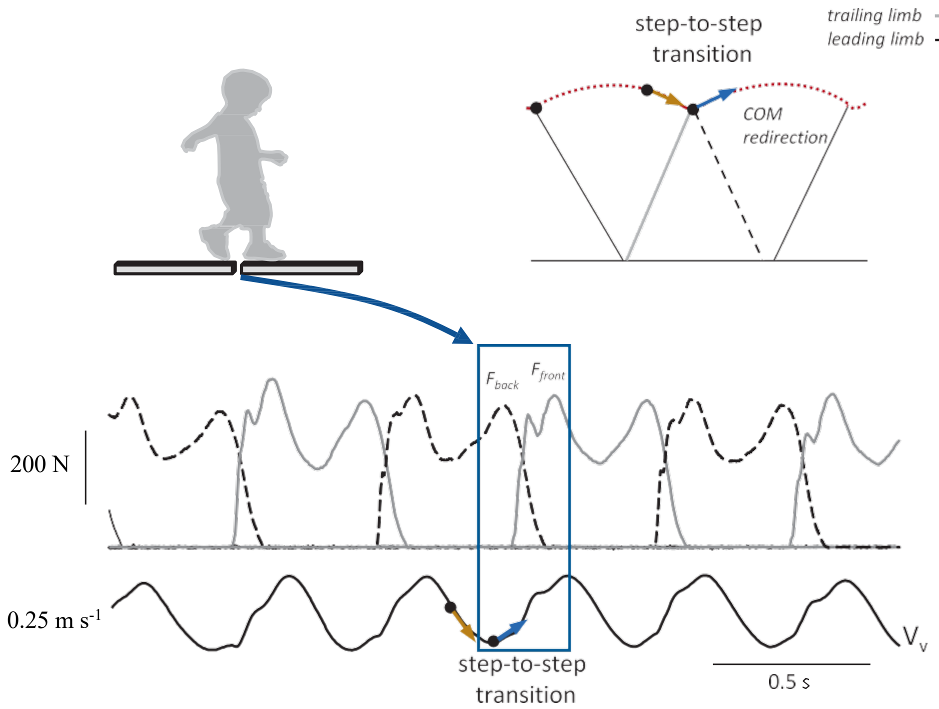
A complete step was selected for analysis only when the feet were on different force plates and when the subject was walking at a relatively constant average height and speed. Specifically, the sum of the increments in both forward and vertical velocity could not differ by more than 25 % from the sum of the decrements (Cavagna et al., 1977). According to these criteria, the difference in the forward speed of the COM from the beginning to the end of the selected step was <6% of the mean speed (except in four instances at very low speeds below 0.56·m·s^{-1}, where it was up to 9%), and the mean vertical force was within 5% of the body weight. A total of 895 steps were analyzed (see Table S1.1). The double contact phase (TDC) was delimited by the front leg foot contact (FC) and the back leg toe off (TO). Froude number ($F_r = v^2g^{-1}l^{-1}$) was measured based on the mean speed (v) and leg length (l) (Table 1). The maximal vertical force exerted by the front leg (F_{front}) and by the back leg (F_{back}) were measured during the step-to-step transition. The ratio between F_{front} and F_{back} was then calculated in order to illustrate the asymmetry in the force generation between the two legs (Dewolf et al., 2022; Meurisse et al., 2019b, 2019a). The vertical velocity of the CoM was determined from the vertical ground reaction force, using the procedure described in detail in Dewolf et al. (Dewolf et al., 2016, 2017). In short, the vertical acceleration of the CoM was computed as $a_v = (F_v - mg)/m$, where m is the subject' s body mass and g is the acceleration due to gravity. The instantaneous velocity in the vertical (V_v) direction was obtained by integration of a_v plus an integration constant (chosen to make the average velocity over a step equal to zero).

Table S1.1 Participants characteristics

Age groups	# of subjects	Age	Leg length	Number of steps analyzed per speed class (m/s)								
(years)	(male)	(years)	(m)	0.2	0.5	0.8	1.1	1.4	1.6	1.9	2.2	2.5
3–4	5 (1)	4.5 ± 0.2	0.49 ± 0.01	2	8	14	20	16	5	0	0	0
5–6	5 (2)	6.2 ± 0.8	0.52 ± 0.01	4	14	20	19	36	14	1	0	0
7–8	5 (1)	8.2 ± 0.4	0.59 ± 0.03	5	10	22	36	23	23	12	8	1
9–10	5 (3)	10.3 ± 0.3	0.65 ± 0.01	0	12	37	42	32	23	27	11	3
11–12	4 (2)	11.9 ± 0.7	0.75 ± 0.02	5	11	21	36	25	33	24	9	0
Adults	12 (6)	22.8 ± 1.9	0.85 ± 0.02	2	18	57	57	70	56	40	19	8

The time of occurrence of the minimal vertical velocity of the CoM ($V_{v,min}$) relative to FC expressed in percent of stride and milliseconds was also calculated. The transition mode was defined by the time of occurrence of the $V_{v,min}$ relative to FC. When the $V_{v,min}$ occurs before FC, the redirection of the CoM velocity was 'anticipated' while when $V_{v,min}$ occurs at or after FC, the redirection of the CoM velocity, and in turn the step-to-step transition, was 'unanticipated' (Dewolf et al., 2022; Franz & Kram, 2013b; Meurisse et al., 2019b).

A.



B. General gait parameters

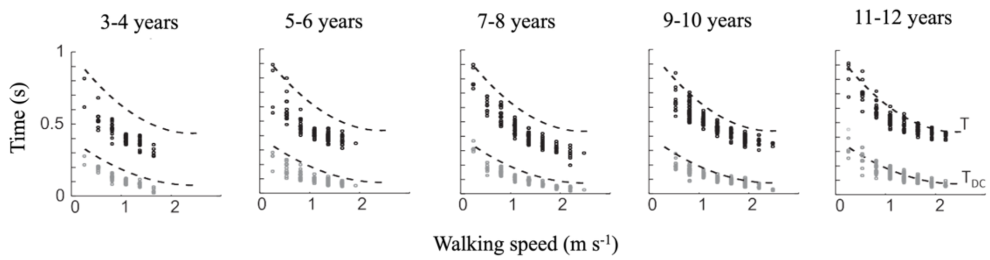


Fig. S1.1. *Typical traces of step-to-step transition in a child of 7 years old and general gait parameters in children and adults.* ***(A)*** *The upper illustration schematically represents the step-to-step transition consequences of walking like an inverted pendulum. The grey and blue arrows indicate the velocity of the CoM before and after its redirection. At the bottom, the curves correspond to the vertical force (F_v) acting upon each leg separately; the continuous grey line is the force back (F_{back}) of the trailing leg; the dotted black line is the force front (F_{front}) of the leading leg as a function of time. The continuous black line below corresponds to the vertical* *velocity (V_v) of the CoM during walking. The blue rectangle delimits one step-to-step transition (from min to max V_v). (**B**) The black and grey dots correspond to the stride period (T) and double contact time (T_{DC}) in children, respectively. The dashed lines represent T and T_{DC} in adults (linear regression as a function of the walking speed). (For interpretation of the references to color in this figure legend, the reader is referred to the web version of this article).*

Results

Temporal parameters

Both walking speed and age ($F_{2,988} = 1416.9$, $p < 0.001$, $R = 0.861$) or Froude Number and age ($F_{2,988} = 1101.5$, $p < 0.001$, $R = 0.831$) statistically predicted stride period. Indeed, the stride period decreased with speed ($p < 0.001$; Fig. S1.1B) and with Froude number ($p < 0.001$) and increased with age ($p < 0.001$). Similar observations were made on the duration of the double stance phase (Fig. 3.1B). Walking speed and age ($F_{2,988} = 1552.6$, $p < 0.001$, $R = 0.871$) and Froude number and age ($F_{2,988} = 1063.4$, $p < 0.001$, $R = 0.826$) statistically predicted the double contact period. The double contact decreased with speed ($p < 0.001$), with Froude number ($p < 0.001$), and increased with age ($p < 0.001$).

Ground reaction force

The Fig. S1.2 presents the differences between the age-groups for the peak of ground reaction forces. The normative data of adults were computed as the linear regression with Froude number as independent variable. The grey area corresponds to the slope plus and minus the 95 % confidence interval. Both walking speed and age ($F_{2,988} = 118.4$, $p < 0.001$, $R = 0.440$) or Froude Number and age ($F_{2,988} = 70.0$, $p < 0.001$, $R = 0.352$) statistically predicted the peak of force developed by the back leg (F_{back}). F_{back} increased with speed ($p < 0.001$) and Froude number ($p < 0.001$) and increased with age ($p < 0.001$). The peak of force developed by the front leg (F_{front}) could also by predicted by speed and age ($F_{2,988} = 1138.9$, $p < 0.001$, $R = 0.835$) or Froude number and age ($F_{2,988} = 1415.4$, $p < 0.001$, $R = 0.861$). F_{front} increased with speed ($p < 0.001$).

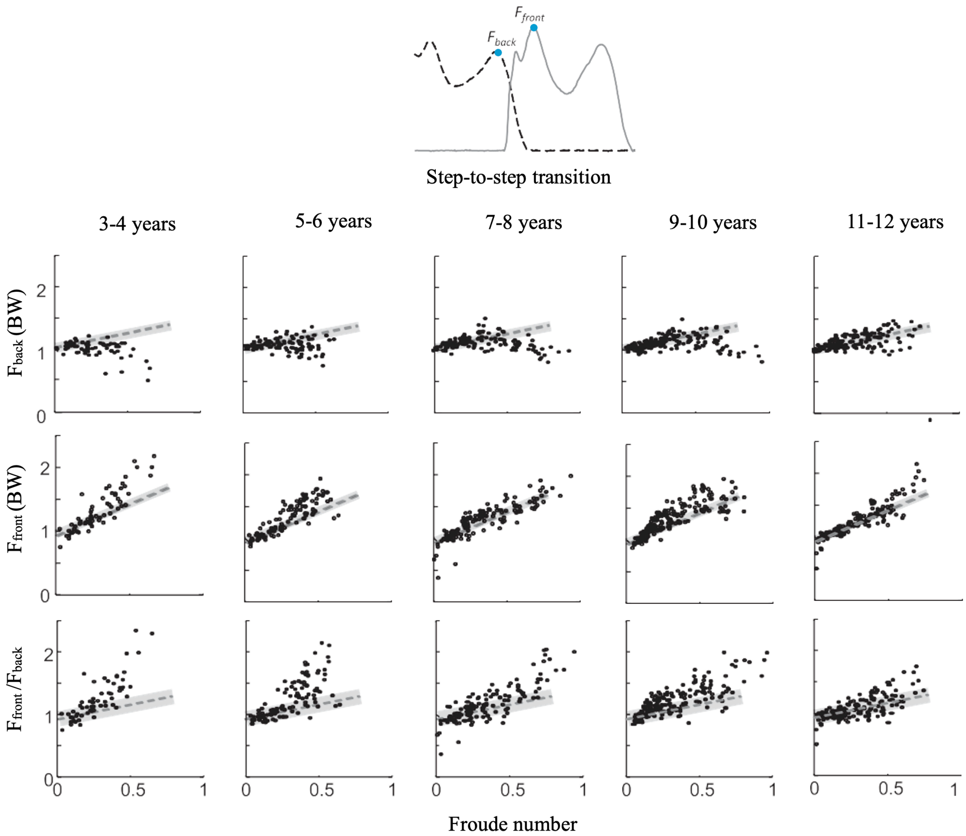
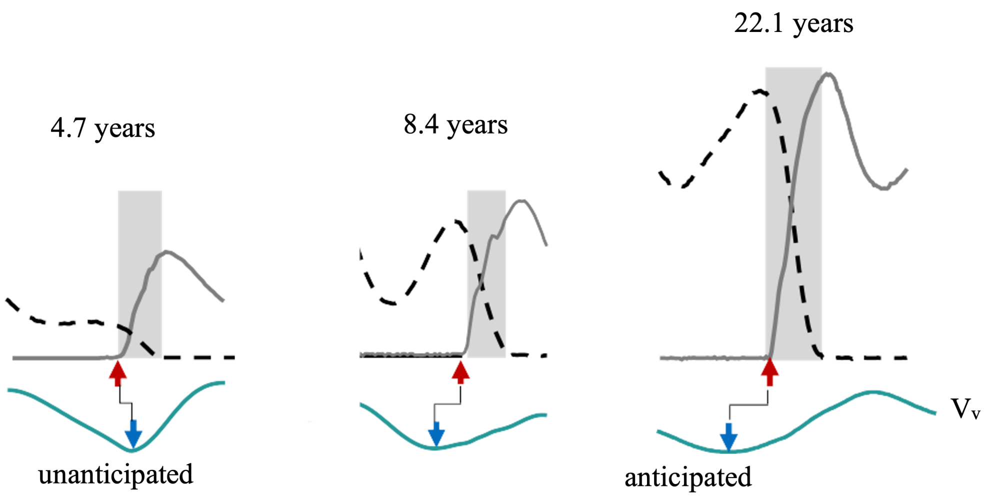


Fig. S1.2. *Vertical ground reaction force under each foot during step-to-step transition. The upper graph illustrates the vertical force exerted by the front (F_{front}) and back (F_{back}) leg during the step-to-step transition (blue rectangle). From top to bottom, the F_{back}, F_{front}, and F_{front}/F_{back} ratio are plotted as a function of Froude number in children (black dots) in each group age (3–4; 5–6; 7–8; 9–10; 11–12 years). The black dashed line and the grey area correspond to the adult normative data. The normative data of adults were computed as the linear regression with Froude number as independent variable. The grey area corresponds to the slope plus and minus the 95 % confidence interval. (For interpretation of the references to color in this figure legend, the reader is referred to the web version of this article.)*

Step-to-step transition mode

The Fig. S1.3B presents the time of occurrence distribution of $V_{v,min}$ relative to foot contact (FC) at all speeds as a function of age-groups in % of stride. Both walking speed and age ($F_{2,988}$ = 117.1, $p < 0.001$, $R = 0.438$) or Froude Number and age ($F_{2,988}$ = 136.2, $p < 0.001$, $R = 0.465$) statistically predicted the delay between $V_{v,min}$ and FC. The delay decreased with increasing speed ($p < 0.001$) and Froude Number ($p < 0.001$). Also, $V_{v,min}$ occurred earlier relative to FC with age ($p < 0.001$). The Fig. S1.3C illustrates the relative percentage of unanticipated transitions (when $V_{v,min}$ occurred after FC) as a function of age-groups and Froude Number. As the delay between the $V_{v,min}$ and FC decreased with Froude number (and speed), the occurrence of unanticipated transition increased, going sometimes up to 100 % of the recorded steps at the fastest speeds. By comparing the younger and older children, it can be seen that this effect of Froude number is dependent on age: in the youngest group, the occurrence of unanticipated transition was higher than in the other group and increased more rapidly with increasing Froude number.

A



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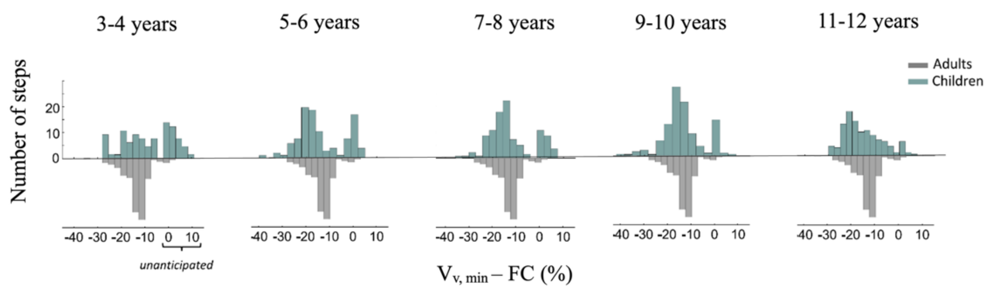

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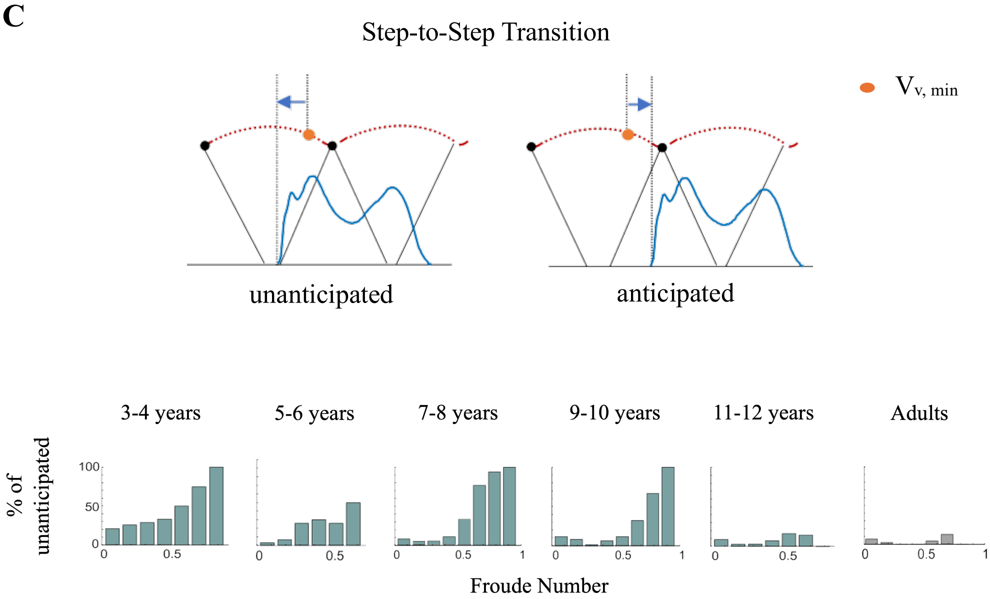

Fig. S1.3. *Anticipated versus unanticipated transition. (**A**) Three representative examples of the force under each foot during a step-to-step transition in 2 children (4.7 and 8.4 years old) and one adult. The bottom curves represent vertical velocity of the CoM (V_v). The red and blue arrows correspond to foot contact and $V_{v,min}$, respectively. (**B**) The bar histograms present the occurrence of the delay between $V_{v,min}$ and foot contact in adults and children (green) (**C**) Schematic illustration of the step-to-step transition mode with representative corresponding vertical forces. The bar graphs below indicate the percentage of unanticipated transition (when the redirection occurs after foot contact) in each age group* *as a function of Froude number. (For interpretation of the references to color in this figure legend, the reader is referred to the web version of this article).*

Discussion

The present study aimed to investigate the step-to-step transition in children younger and older than 7 years old and compare it with adults. To do that, we measured the vertical ground reaction forces under each foot during a walking step, compared the force exerted by the leading and trailing limb during the weight transfer from one foot to the other (Fig. S1.3A), and classified the transition as originally proposed by Meurisse et al. 2019 (Dewolf et al., 2022; Meurisse et al., 2019b) in older adults: when the down-to-up redirection of the centre of mass of the body (COM) begins during the double stance, the transition is considered as non- anticipated whereas the anticipated transition mode occurs when the redirection begins before the contact of the leading leg, initiated by the vertical push of the trailing leg. According to Meurisse et al. (Meurisse et al., 2019b), the transition mode is dependent on speed. Therefore, we also analyzed the transition at different speeds and corresponding Froude numbers (see Table S1.1). Our findings revealed a higher rate of non-anticipated transition modes in younger infants and at fast walking speed (Fig. S1.3).

The modified transition is related to a different distribution of vertical ground reaction forces between the leading and the trailing limb (Fig.S1.2), with a reduced propulsion of the trailing limb compensated by a higher peak of force under the front foot. Some previous study has suggested that after the age of 4 years old, step parameters are already ‘adult-like’ (Sutherland, 1997). More recently, it has been suggested that, in terms of center of mass dynamics and lower-limb kinematics, gait maturity is reached around 7 years of age (Cheron, Bengoetxea, et al., 2001; Cheron, Bouillot, et al., 2001; Dominici et al., 2010; Ivanenko, Dominici, et al., 2004; Ivanenko et al., 2005, 2007). By comparing children younger and older than 7 years old to adults, the results of the present study indicate that the gait is not yet fully mature at the age of 7 years old. Indeed, both younger and older children do not anticipate the foot contact and fail to redirect of the CoM motion before double stance, as adults do. While it should be taken into account that the degree of gait maturity does not always relate directly to the chronological age of the child (Bach et al., 2021), the difference with mature gait was, however, greater in the younger groups. One possibility is that peripheral, biomechanical factors may play the critical role in determining why children walk the way they do. For example, strength and balance are keystones of improvements in walking skill (Backman et al., 1989), and the deviation from adult gaits in infancy may also be related to adaptive strategies for limiting the muscle activation demands (Hubel & Usherwood, 2015). This is also supported by the lack of power generated at the ankle joint before foot-off (Forssberg, 1985; Hallemans et al., 2006).

Another possibility is that practice is the critical factor for the developmental course of walking. In fact, the constant learning and exploration of movement patterns in children suggest that motor loco- motion patterns are not fixed but are flexible. The progressive transition to anticipated transition, as adult do, can be related to the acquisition of the adult-like heel-to-toe rolling pattern of the foot (A. H. Dewolf et al., 2020; Hallemans et al., 2006) and the acquisition of the adult-like mechanics and exchange between potential and kinematic energy of the CoM when walking in the gravity field (Cavagna et al., 1983). From a metabolic point of view, since step-to-step transition is major determinant of the overall cost of walking (Kuo et al., 2005), the higher energy cost of walking observed in children until the age of 10 years old (DeJaeger et al., 2001) may be also explained by the greater collision at foot contact than adults at the same speed (Fig. S1.2).

Interestingly, similar results were observed in older adults. Indeed, a greater contribution of the leading limb compensating a reduced vertical support of the trailing leg has been observed. As a result, the transition is initiated later in the gait cycle with age (Dewolf et al., 2022; Meurisse et al., 2019b, 2019a). Also, the late CoM redirection was associated with age-related decline of gait, such as poorer static balance, modification of limb coordination, greater mechanical work, and differential organization of spinal motor output (Dewolf et al., 2022). Based on the previous consideration, one may expect that the step-to-step transition in children as in older adults may reflect a deviation from adult mature gait.

Conclusion

In conclusion, our results show that the rate of unanticipated transition decreased with age, as speed and Froude number increased. Because the modification of step-to-step transition has been related to the modification of neuromuscular control of gait, assessing step-to-step transition in clinical practice could be used as a quantitative assessment of the development of locomotion.

Acknowledgments

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Supplementary Chapter II

Unraveling age-related impairment of the neuromuscular system: exploring biomechanical and neurophysiological perspectives

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Keywords: neuromechanics, gait, physio-mechanics, direct current spinal stimulation, walking pattern

List of Abbreviations

tsDCS:	Trans-spinal direct current stimulation
PICs:	Persistent inward currents
MN:	Motoneuron
EMG:	Electromyography
CoM:	Center of mass
GRF:	Ground reaction force
PCA:	Principal component analysis
H-reflex:	Hoffmann reflex

Abstract

Background: With extended life expectancy, the quality of life of older adults is a priority. Loss of mobility, increased morbidity and risks of falls have dramatic individual and societal impacts. Here we consider age-related modifications of gait, from a biomechanical and neurophysiological perspective. Among the many factors of frailty involved (e.g., metabolic, hormonal, immunological), loss of muscle strength and neurodegenerative changes inducing slower muscle contraction may play a key role. We highlight that the impact of the multifactorial age-related changes in the neuromuscular systems results in common features of gait in the immature gait of infants and older adults. Besides, we also consider the reversibility of age- related neuromuscular deterioration by, on the one hand, exercise training, and the other hand, novel techniques such as direct spinal stimulation (tsDCS).

Introduction

With extended life expectancy, the quality of life of elders is a priority. Loss of mobility, increased morbidity and risks of falls have dramatic individual and societal impacts. Among the many factors of frailty involved, loss of muscle mass and strength (Akima et al., 2001) and neurodegenerative changes (Rygiel et al., 2016a) play a key role. Whether changes in the neural control precede or follow the decline of muscle mass and strength and how they both are related to gait alteration remains yet to be established. More than ever, this needs to be elucidated to implement interventions that can maintain or improve neuromuscular function in older adults.

Biomechanical changes with age have garnered considerable scientific attention for nearly 50 years. The scientific community (Delabastita et al., 2021; Winter, Patia, et al., 1990) most often points to a reduction in mechanical power generated by the plantar flexor muscles during the push-off phase of walking as the hallmark biomechanical ageing features of gait. However, the 11%–35% decline in force or power-generating capacity of propulsive leg muscles cannot fully explain the age-related modification of gait in older adults. Indeed, (i) many old adults underutilize their available muscular capacity for generating propulsive power in walking and are able to increase it during slope walking or using biofeedback (Browne & Franz, 2019) , and (ii) age-related changes in kinematics have been found prior to the appearance of propulsion decline with increasing age (Sloot et al., 2021). Taken together, it suggests that the decline of propulsive power generation is thus not only due to a reduced muscular capacity, but neural factors are likely to contribute as well. For instance, with aging motor weakness is due in part to neuromuscular degeneration, but also to degenerative changes in the central nervous system. Thus, reduction in grey matter volume (Good et al., 2001), number of motor cortical (Henderson et al., 1980) and spinal motor neurons (Doherty, 2003), synaptic density (Haug & Eggers, 1991), white matter integrity (Davis et al., 2009), and descending commands for motor activation (Yue et al., 1999) are some of the factors that may contribute to age-related motor impairment. Another determinant of functional capacity and autonomy is the integrity of other components of the neuromuscular system, which wires the brain and skeletal muscles via motor neurons and the neuromuscular junction. However, despite its obvious importance for rhythm generation, the potential involvement of the spinal cord in age-related modification of locomotion has received little attention.

Using an electrophysiological approach, a way to get insight into spinal cord functioning is to look at the spatiotemporal organization of the total locomotor output by mapping multi-muscles EMG onto the spinal cord in approximate rostro-caudal locations of the motoneuron (MN) pools (Cappellini et al., 2010; A. H. Dewolf, Ivanenko, et al., 2019a; Ivanenko et al., 2008, 2013; La Scaleia et al., 2014; Yokoyama et al., 2017). By studying the spinal motor output across various walking conditions in older adults (walking at different speeds, backward, upslope, downslope, upstairs, downstairs), similar age-related differences in muscle activations have been observed despite the various biomechanical constraints (Dewolf et al., 2021a, 2021b). In particular, the activity profiles of the muscles innervated from the sacral segments were significantly wider in older adults in all conditions. Interestingly, similar modification has been observed in young children (A. H. Dewolf et al., 2020; Ivanenko et al., 2013).

The major consideration of this review is the age-related remodeling of both the neural and muscular system and its relationship with locomotion changes with age, to shed light on the multifactorial age-related changes of gait. The alterations of the gait pattern in older adults are then compared to immature gait. Besides, we also consider the reversibility of age-related neuromuscular deterioration by, on the one hand, exercise training, and the other hand, novel techniques such as direct spinal stimulation (tsDCS) to mitigate the reduction of intrinsic spinal motoneuron excitability in older adults (Orssatto, Borg, et al., 2021), and how it could potentially lead to improved strategies for promoting locomotor function recovery.

Neuromuscular modification with aging

Aging is a natural and gradual process where the alterations in motor control and physical fitness are multifactorial (Borzuola et al., 2020). Strength capacity and muscle mass decrease during aging, in great part due to sarcopenia. Aging-related sarcopenia is the most common type of atrophy in humans. Specifically, sarcopenia is a progressive skeletal muscle disorder identified by low muscle strength, low muscle quantity or quality, and low physical performance (Cruz-Jentoft et al., 2019). It is associated with an increased risk of adverse outcomes, such as functional disability, poor quality of life, and a higher risk of mortality (Batsis et al., 2014; Beaudart et al., 2017; Kelley & Kelley, 2017; Liu et al., 2017). In older adults, the loss of strength seriously affects independence associated with activities of daily living but also leads to a greater risk of falls, which is strongly related to mortality (Landi et al., 2012; Suzuki et al., 1992). From a clinical perspective, it is essential to understand the mechanisms underlying the modifications in skeletal muscle morphology and function, which are evident during aging (Fiatarone, 1990). Muscle strength begins to decline after 30 years of age and continues to decline with advancing age (Gava et al., 2015). Changes related to muscle morphology and its electrophysiology generally appear after the age of ~40, and it also continues to decrease progressively (Murton, 2015; Oertel, 1986; Stalberg & Fawcett, 1982). Therefore, changes in strength appear to precede changes associated with skeletal muscle morphology. Also, the effect of aging on skeletal muscles depends on muscle location and function, since leg muscles are more affected than arm muscles (Oertel, 1986; Stalberg & Fawcett, 1982).

Several age-related modifications of muscle tissue have been described, such as loss of muscle fibers (Lexell, 1995; Lexell, Henriksson-Larsén, et al., 1983; McPhee et al., 2016) or substantial loss of contractile proteins (Larsson et al., 1995), such as myosin heavy chain (Siparsky et al., 2014). Not only the reduction in the total number of fibers occurs, but also in their cross-sectional area (Lexell & Taylor, 1991b). It should be noted that a differential response is reported in fiber loss depending on the type of muscle, with a faster decrease mainly observed in type II fibers (Cade & Yarasheski, 2006; Domingues-Faria et al., 2016). Also, a change from the fast myosin isoform to the slow isoform has also been observed, which has a lower capacity to generate force. This change in fiber type could contribute to both slowing movement and decreased maximal strength, and in turn induce age-related changes of gait.

Even the pathogenesis of sarcopenia is not yet fully understood, multiple etiological factors seem to be involved, including alteration of muscle proteostasis (Lecker et al., 2004), mitochondrial dysfunction and mitochondrial DNA deletions (Bua et al., 2006), deregulation of satellite cells (Shefer et al., 2006), and accumulation of extracellular matrix called fibrosis and fat infiltration into skeletal muscle (M.-Y. Song et al., 2004). The cause of sarcopenia cannot be solely attributed to alterations in skeletal muscles. In fact, the nerve responsible for stimulating muscle fibers plays a significant role in sarcopenia. As the skeletal muscles experience degeneration with age, the decline in neuromuscular function emerges as a crucial contributing factor (Aagaard et al., 2010). Evidence has shown alterations associated with a reduction in motor units. Studies have specifically compared the amount of motor neurons in young and elderly subjects, with the latter showing a 50% decrease (Doherty et al., 1993). In addition, the motor neurons begin to exhibit alterations in firing frequency and rate. The maximum firing frequency of motor neurons is lower compared to young subjects (Klass et al., 2008). Other changes have also been detected during aging related to a decrease in axonal conduction velocity, which is explained by reduced myelination and internodal length (Scaglioni et al., 2002). Therefore, evidence suggests that the compromised nervous system function may also be one of the important contributors to functional decline described in sarcopenia (Kwon & Yoon, 2017; Rygiel et al., 2016a). Indeed, normal innervation and its corresponding regular activation are necessary to maintain muscle mass through muscle contraction. For example, there is an association between the loss of muscle fibers and the loss of motor units in older people (McNeil et al., 2005; Piasecki et al., 2016, 2018). In addition, slow motor neurons may be more adapted to reinnervation, leading to the loss of fast motor neurons with age. This could respond to the change in fiber type that occurs with aging (Andersen, 2003; Kadhiresan et al., 1996; Larsson et al., 1978).

While the causes of the age-associated loss of motor neurons are still unsettled, the neuromuscular junction integrity, and in particular the mitochondrial dysfunction at the neuromuscular junction, may have an important role (Shigemoto et al., 2010). The changes in the neuromuscular junction have been reported to be related to morphological alterations of the pre- and post-synaptic regions and to the reduction of synaptic vesicles (Jang & Van Remmen, 2011). The loss of motoneurons also plays an important role in the alterations of the excitation-contraction coupling process during aging (A. M. Payne & Delbono, 2004). Indeed, the decrease in isometric strength and contraction velocity appears before the reduction in muscle mass. Therefore, it has been proposed that the decrease in the number of motor units occurs before the loss of muscle function (Deschenes et al., 2010; Sheth et al., 2018). However, there is still insufficient evidence, and more studies are required to complement the current hypothesis.

Gait during development and aging: a brief overview of the two sides of life

The multifactorial age-related changes in the neuromuscular system, summarized in the last section, are rather well documented. As people age, those changes result in alterations in gait patterns. In this section, we present different aspect of gait (presented in Fig. 2.1) that are affected by age, and that interestingly resemble those seen in younger infants.

In humans, when EMG activity patterns are mapped onto the spinal cord in approximate rostro-caudal locations of the motoneuron (MN) pools, the activation of MNs tends to occur in bursts that can be associated with the major kinetic events of the gait cycle (Cappellini et al., 2010; A. H. Dewolf, Ivanenko, et al., 2019a; Ivanenko et al., 2008; La Scaleia et al., 2014; Yokoyama et al., 2017). In particular, the first burst occurs around foot contact and is mainly localized on the lumbar segment whereas the second burst occurs during the second part of the stance mainly and is localized on the sacral segment (Figure 1A). This approach provides information about pattern output in terms of lumbosacral segmental control (from L2 to S2) rather than in terms of individual muscle control.

In older adults, across different forms of walking gait, age-related differences were observed (Dewolf et al., 2021a, 2021b; Monaco et al., 2010; Santuz et al., 2020), suggesting specific adjustments of the pattern generation circuitries. In particular, the sacral output was significantly wider in older adults and occurred earlier in the stance (Figure 1A). Interestingly, this result does not simply reflect the documented distal-to-proximal modification of kinetics since the human spinal topography does not reflect the muscle topography on the lower limbs (F. P. Kendall et al., 2005). Instead, this potentially highlights a distal to proximal degeneration of the motor system. Accordingly, when the spinal excitability is estimated using the Hoffmann reflex technique, no difference is found between young and older adults on vastus medialis muscle (Mau-Moeller et al., 2013), whereas age-related modulations of the reflex response have been reported in soleus muscle (Baudry et al., 2015). Interestingly, while the craniocaudal gradient of corticospinal development in infancy is well established (G. V. Payne & Larry D., 2017), less is known about the differential degeneration of different portions of the corticospinal tract with aging.

Development and aging can be seen as two opposite but complementary phenomena (Feltes et al., 2015). For example, it appears that projection tracts, such as the corticospinal tract, which develop earlier than association tracts in infancy, degenerate later than association tracts in older subjects. Also, primitive reflexes, which are commonly present in normal infants and disappear during development, reappear in patients with diseases of the nervous system but also in healthy older adults with an incidence increasing with age (Damasceno et al., 2005; Douglas Gossman & Lawrence Jacobs, 1980; Hobo et al., 2014; Van Boxtel et al., 2006). Indeed, attempts to elicit primitive reflexes are a routine part of the standard neurological examination in the elderly, with the following reflexes tested: e.g., snout, suck, palmomenral, and hand grasp (described in detail by Koller (Koller, 1984)). Another reflex observed at the beginning of life is the stepping reflex: human newborns step on the ground if supported (A. H. Dewolf et al., 2020; Dominici et al., 2011; Forssberg, 1985; Thelen & Fisher, 1995; Yang et al., 1998), and stepping generally disappears a few weeks after birth unless trained. The relationship between this reflex and mature walking gait has been argued (Andre-Thomas & Autgaerden, 1966; Dominici et al., 2011; Sylos-Labini et al., 2022), with lower ow complexity and higher variability of neuromuscular signals in neonates. Because of the less complex and more variable control of muscle in older adults (J. L. Allen & Franz, 2018), and based on the common features of gait between infants and older adult presented in the present section (Figure 2.1), One may speculate about the potential greater similarities between neonatal stepping and older adult's gait pattern. Also from a kinematic point of view, a simpler coordination pattern among the lower limb segments can be observed both during childhood and agedness (Bleyenheuft & Detrembleur, 2012; A. H. Dewolf, Ivanenko, et al., 2019a; Dominici et al., 2010; Gueugnon et al., 2019b; Ivanenko, Dominici, et al., 2004; Noble & Prentice, 2008). One way to unravel the multi-segmental coordinative law is the so-called coplanar variation (Bianchi, Angelini, Orani, et al., 1998; Borghese et al., 1996). During walking, each lower-limb segment oscillates back and forth relative to the vertical with a similar waveform, time-shifted across different segments (Figure 1C). The lower limb segment angles do not evolve independently of each other, but they are tightly coupled: when plotted one vs. the others, they co-vary along a plane, describing a characteristic loop over each stride (Figure 2.1C). The specific shape and orientation of the plane reflects the phase relationship between segments and therefore the timing of the intersegmental coordination (Barliya et al., 2009). Even if the intersegmental coordination in toddlers rapidly evolves toward the adult shape with experience (Cheron, Bouillot, et al., 2001; Dominici et al., 2010; Ivanenko, Dominici, et al., 2004), when toddlers step in various conditions (slope, stairs, backward), they do not adapt their segmental coordination as adults do. Instead, they keep constant phase relationships (Dominici et al., 2010) (Figure 2.1C). In older adults, the modification of plane orientation across gait conditions is less adapted than in young adults (Dewolf et al., 2021a, 2021b, 2019a). Since the changes in planar covariation are thought to reflect the ability to adapt to different gait conditions (Bianchi, Angelini, & Lacquaniti, 1998; A. H. Dewolf et al., 2018; Martino et al., 2014), the lack of changes observed in toddlers and to a lesser extent in older adults suggest reduced ability to adapt gait to environment or specific constraints (Dominici et al., 2010).

Because a link between center of mass (COM) trajectory and functional spinal cord topography has been previously highlighted (Cappellini et al., 2010; A. H. Dewolf et al., 2020; A. H. Dewolf, Ivanenko, et al., 2019a), one may expect comparable COM dynamics in older adults and young infants. Center of mass (COM) mechanics is a fundamental concept in biomechanics that describes the movement and balance of an individual's body. In young adults, during walking the COM vaults over a relatively stiff limb with the heel well in front of the hip at the beginning of the stance, and the heel lift with maintained toe contact at the end of the stance. One of the direct consequences of such a heel-to-toe roll-over pattern is that the extension of distal joints is delayed relative to proximal joints, leading to the typical double hump shape (so-called ≪m − pattern≫) of the vertical ground reaction force (Hallemans et al., 2006) (Figure 2.1A). In both older adults and younger infants, the walking gait lacks the specific m-pattern shape of adult heel-to-toe roll-over walking pattern (Dominici et al., 2011; Forssberg, 1985; Sylos-Labini et al., 2017), due to the lack of late push-off from the trailing leg (A. H. Dewolf, Ivanenko, et al., 2019a; Dominici et al., 2011; Gueugnon et al., 2019b). Another similarity is that both young infants and older adults have limited control over their COM (Foster et al., 2019; Malloggi et al., 2019). In the next section, we discussed the potential cause of modification of gait in older adults, resulting in kinematics, kinetics and neural similarities with the gait observed in children.

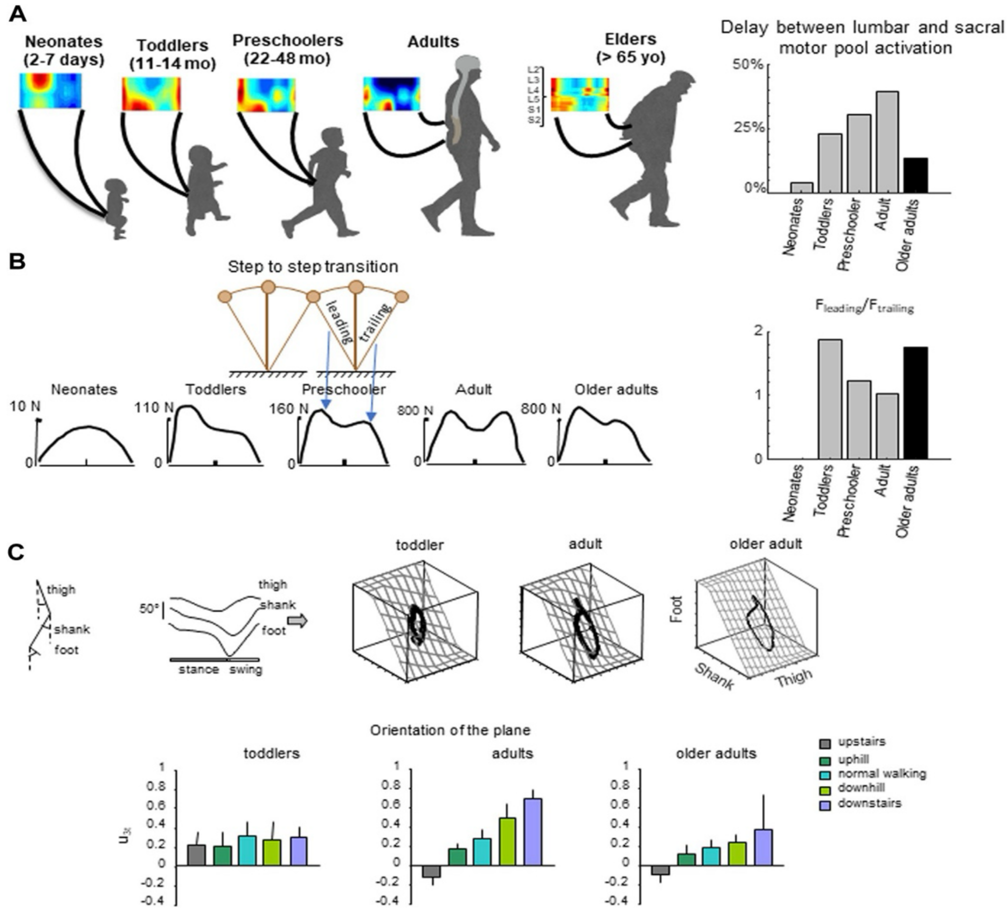

Fig. S2.2.1: *General features of gait in older adults. (A) spatiotemporal maps of motoneuron activity of the lumbosacral enlargement in neonates, toddlers, preschoolers, adults and older adults, and the delay between the activation of lumbar and sacral activation (data from Ivanenko et al. (2013) for neonates, toddlers, preschoolers and adults, and from Dewolf et al. (2021a) for older adults). (B) Schematic representation of walking like an inverted-pendulum. Below, representative vertical loading force during stepping in neonates, toddlers, preschoolers, adults and in older adults. The characteristic force profile was evaluated using the ratio between the leading limb and the tailing limb. (C) Intersegmental coordination assessed by principal component analysis (PCA) of limb segment elevation angles during walking. From left to right: thigh, shank, and foot elevation angles (relative to the vertical), corresponding 3D trajectory in segment angle space along with the interpolated plane (modified from Ivanenko et al. (2008)). Three examples of gait loops are presented (one toddler, one adult and one older adults). Below, changes in the orientation of the covariance plane during walking over different surfaces in toddlers, adults and older adults (modified from Dominici et al. (2010) and from Dewolf et al. (2021b)).*

Could we counteract the age-related modification of gait?

Based on the well-documented change in muscle strength with aging described in the last section, a lot of efforts have been made to counteract it using exercise (Christie, 2011; Hortobágyi et al., 2015; Keating et al., 2021; Wu et al., 2021). Physical training is reported as an effective treatment for maintaining muscular function (Chen et al., 2021; El Hadouchi et al., 2022; Kraemer, Ratamess, et al., 2002; Markov et al., 2023). Guidelines recommend high physical activity levels to increase health benefits in older adults (Boyer et al., 2012; D. Taylor, 2014), since it is supposed to enhance daily activities like gait. Resistance or power training not only increases/maintains muscle mass, strength, power and functional capacity in older adults, but it also induces several neuromuscular adaptations, such as an increase in peak firing frequencies of motoneurons. Also, older adults still practicing long-distance running reduce the decline in muscle function with age by enhancing the neural drive to the muscle (Cogliati et al., 2020).

Resistance training (RT) positively affects walking speed (Keating et al., 2021). For example, Hortobágyi et al. (2015) found that RT significantly increases the habitual gait speed of healthy old adults by 8.4% as a long-term effect. Power training also impacts gait velocity (Beijersbergen, Granacher, Gäbler, DeVita, et al., 2017; Beijersbergen, Granacher, Gäbler, Devita, et al., 2017a, 2017b), changing the rate of force development, which is moderately correlated with gait speed (Stock et al., 2019), and improving the functional performance (Radaelli et al., 2023). The main related effect of muscular training is a higher 'habitual walking speed' after the exercise sessions. However, training fails to directly translate to improved propulsive power generation in walking (Beijersbergen et al., 2013). For example, a higher level of physical activity in older adults did not mitigate the age-related modification of kinematic coordination and distal-to-proximal redistribution (Boyer et al., 2012). Also, greater muscular power (more than muscle strength) has been reported to have a strong influence on mobility (Bean et al., 2003), but without a clear change in gait pattern. Indeed, the biomechanical, physiological, and motor control adaptations in gait with training are still unknown, and physical trainers or physiotherapists lack consistent biomechanical data to understand the adaptation mechanism (Beijersbergen, Granacher, Gäbler, Devita, et al., 2017b). Based on lower limb coordination, Bianchi et al. (Bianchi, Angelini, & Lacquaniti, 1998) showed that trained young subjects can exploit better the dynamic coupling between segments to save mechanical energy than untrained young subjects. It is, therefore, plausible that training in older adults may affect the lower limb intersegmental coordination to allow optimized gait mechanics.

While walking speed has been suggested to predict frailty and disability in older adults (Guralnik et al., 2000), we believe that evaluation of spontaneous walking speed is not the best outcome to evaluate the age-related decline of gait. Spontaneous gait speed, if not performed after period of familiarization sessions and following standardized instructions, may vary with the mood, motivation, stimuli of the experimenters, etc. For example, in a classical paper, Bornstein and Borstein (Bornstein & Bornstein, 1976) showed that the 'pace of life', measured as the spontaneous speed, varies with the size of the local population, regardless of the cultural setting, suggesting that immediate social and physical environment exert strong control over individual habitual speed (Levine & Norenzayan, 1999). Therefore, we believe that there's an imperative need to understand the role of physical exercise in the process of age-related modification of neuromuscular control of gait. In particular, the data to interpret the mechanism needs to be more quantitative.

Enhancing physical capacity alone may not be sufficient to mitigate the age-related decline of the neuro-muscular system, such as the distal to proximal degeneration of the motor system highlighted above. A recent rehabilitation approach is the use of real-time biofeedback to encourage favorable biomechanical adaptations. For example, it has been showed that the propulsive power can be increased during walking in older adults using ankle power biofeedback (Browne & Franz, 2019), resulting in a reduction of distal-to-proximal redistribution of joint efforts. Based on the age-related modification of gait highlighted in Fig. 1, one may expect that other parameters, such as the center of mass trajectory, can be manipulated using real-time biofeedback (e.g., as in (Massaad et al., 2007)). In particular, using real-time biofeedback to cue an acute change in the peak of vertical ground reaction force or in the limb loading may be an effective gait training intervention to mitigate the effect of age in older adults. Such approach has been developed in gait retraining following anterior cruciate ligament reconstruction (Armitano-Lago et al., 2020; Luc-Harkey et al., 2018) but not, to the best of our knowledge, with older adults.

Also, not only muscles but also the firing characteristics of our spinal motoneurons play a critical role in producing force, and so, performing daily activities. Motoneuron firing is determined by complex factors, such as ionotropic synaptic input and persistent inward currents (PICs) (Orssatto, Mesquita, et al., 2021). PICs are depolarizing currents generated by voltage-sensitive sodium and calcium channels. Hassan et al. (Hassan et al., 2021b) (2021) found weaker estimates of PICs in older adults than in their younger counterparts, and propose that this weakening is an underlying mechanism for the slowing of motoneuron firing with ageing. Interestingly, the similarities observed between infants' and older adults' locomotor patterns (Figure 1) might be related to the slower and weaker firing characteristics (Dayanidhi et al., 2013).

As described by Hassan et al. (Hassan et al., 2021b), the PICs weakening might result from a multitude of factors: (1) deterioration within the monoaminergic systems, (2) imbalance between excitatory and inhibitory synaptic inputs, or (3) changes in the function of monoaminergic receptors or voltage-gated channels. The question that need to be answered is now: how can we counteract the age-related decline in PICs? Our proposed answer for a future research question in this context is spinal neuromodulation, a promising strategy to augment spinal cord activity. In particular, non-invasive trans-spinal cord direct current stimulation (tsDCS) may improve spinal motor circuit function and motor output (Jankowska, 2017; W. Song & Martin, 2017) in older adults, because of the increase in firing frequencies of motoneuron (Baczyk, M., 2019) its specific effect of augmenting PIC-like responses induced by c-tsDCS, L-type Ca2C channel activation (W. Song & Martin, 2017).

The tsDCS has been increasingly used over recent years in the rehabilitation of patients following neurological injuries (Gad et al., 2021; Ievins & Moritz, 2017; C. Taylor et al., 2021) or as an addition to physical training in sports (Berry et al., 2017). The effect of tsDCS on the gait patterns of older adults has not been studied yet. However, enhancing the PIC-like response of motor units would be well-suited to mitigate the effect of aging on spinal motor output. We hope that the ideas presented here help to motivate future efforts in understanding the quantitative modification of gait with aging and in evaluating a promising method that could be used as a supplementary tool in the management of geriatric patients.

Concluding remarks

This review outlines great similarities between the 'first steps' of infants and the 'last steps' of older adults. While part of the modifications observed in older adults may emerge from a lack of propulsive power, other neurodegenerative changes play a key role. In particular, slower muscle contraction is observed, resulting from the change in fiber type, the greater reinnervation of slow motor neurons or the lower motoneuron firing frequency with ageing, which is also an important peripheral contributor to the lack of adult-like locomotor patterns in early infancy (Dewolf et al., 2020). Gaining insights into the age-related changes in human gaits may provide important clinical implications. For instance, we propose a novel intervention to enhance the PIC-like response of the motor unit, and in turn, mitigate the effect of aging.