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Age-related changes in muscle fatigue and force control of the trunk extensor muscles

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ABBREVIATIONS

ANOVA	Analysis of Variance
ART	Aligned Rank Transform
ATP	Adenosine Triphosphate
AXIS	Appraisal tool for Cross-Sectional Studies
BBS	Berg Balance Scale
BMI	Body Mass Index
BPAQ	Baecke Physical Activity Questionnaire
CEDE	Consensus for Experimental Design in Electromyography
CI	Confidence Interval
CoV	Coefficient of Variation
CSA	Cross-Sectional Area
EMG	Electromyography
sEMG	Surface Electromyography
ES	Erector Spinae
FCSA	Functional Cross-Sectional Area
FZ	Fisher’s z values
GLMMs	Generalised Linear Mixed Models
GRADE	Grading of Recommendations Assessment, Development and Evaluation
H-reflex	Hoffman reflex
HDsEMG	High-density Surface Electromyography
IQR	Interquartile Range
LBP	Low Back Pain
LES	Lumbar erector spinae
LMMs	Linear Mixed Models
MD	Mean Difference
MeSH	Medical Subject Headings
MF	Median Frequency
MFD	Multifidus
MFCV	Muscle Fibre Conduction Velocity
MPF	Mean Power Frequency
MSC	Magnitude Squared Coherence
MVC	Maximal Voluntary Contraction

MVIC	Maximal Voluntary Isometric Contraction
MVIF	Maximal Voluntary Isometric Force
NMJ	Neuromuscular Junction
OLDER	Older adults
PA	Physical Activity
PCA	Principal Component Analysis
PICOS	Population, Intervention, Comparison, Outcomes and Study design
PRISMA	Preferred Reporting Items for Systematic Reviews and Meta-Analyses
PROSPERO	International Prospective Register of Systematic Reviews
REML	Restricted Maximum Likelihood Estimation
RMS	Root Mean Square
SD	Standard Deviation
STROBE	Strengthening the Reporting of Observational studies in Epidemiology
WHO	World Health Organization
YOUNG	Young adults

ABSTRACT

The overall aim of the three studies presented in this PhD thesis was to enhance our understanding of age-related adaptations of the trunk extensor muscles, with a specific focus on endurance capacity, muscle force control (torque steadiness), and the potential underlying neuromuscular mechanisms by using high-density electromyography (HDsEMG).

The *first study* (**Chapter 2**) examined whether the spatial distribution of lumbar erector spinae (LES) activity, assessed using HDsEMG, differs between younger and older individuals in the presence of muscle fatigue. The results demonstrated age-related differences in LES activity adaptations to fatigue, with older adults exhibiting a cranial shift in the centroid of muscle activity, unlike younger controls. This likely reflects a protective, yet less efficient, motor control strategy that is adopted with ageing, potentially contributing to the greater force fluctuations observed in older adults during the fatiguing task.

The *second study* (**Chapter 3**) was a systematic review and meta-analysis evaluating current evidence on how ageing influences spinal extensor muscle fatigue. The results of the meta-analysis revealed significantly reduced endurance time of the back extensor muscles during isometric tasks in older adults compared with younger controls. However, inconsistencies in EMG findings across studies limit the understanding of the neuromuscular mechanisms underlying this age-related decline.

The *third study* (**Chapter 4**) examined force control of the trunk extensor muscles during isometric and isokinetic concentric contractions, focusing specifically on torque steadiness and HDsEMG-torque coherence. The results revealed that age-related impairments in torque steadiness were greater during isokinetic than isometric trunk extension contractions, especially at the lower intensity (25% of maximal voluntary contraction). In addition, distinct neuromuscular patterns were observed: older adults showed reduced HDsEMG-torque coherence magnitude during isokinetic contractions and an altered spatial distribution of coherence in the LES muscles during isometric tasks. Collectively, our findings suggest that these altered neuromuscular patterns may limit optimal force production in older adults.

CHAPTER 1

General introduction

This chapter introduces the key themes of the thesis by reviewing the literature on ageing, muscle fatigue, and force control, specifically in relation to the trunk extensor muscles. It also presents high-density surface electromyography as a methodological tool for investigating neuromuscular function in this context.

1.1 Ageing

1.1.1 General overview

Ageing is a multifactorial and progressive biological process characterised by a gradual decline in physiological functions and a loss of biological integrity, ultimately increasing vulnerability to disease and mortality (Vasto et al. 2010). However, these changes are neither linear nor uniform, as the rate and expression of normal ageing can vary significantly between individuals. This variability results from complex interactions between genetic factors, environmental exposures, and behavioural traits (Castruita et al. 2022).

Ageing is a global phenomenon with profound implications for healthcare systems, social services, and economies, as the proportion of older adults in the population continues to increase worldwide. Conventionally, the term “elderly” refers to individuals aged 65 years and older. Within this group, those aged 65 to 74 are classified as the “early elderly,” while individuals aged 75 and above are referred to as the “late elderly” (Orimo et al. 2006). However, the definition of older adults can vary depending on the country, socio-cultural context, and life expectancy. While the age of 65 and above is commonly used in most developed countries, the United Nations typically adopts a lower general threshold, defining older individuals as those aged 60 years and above. In regions where life expectancy is lower, the threshold can be set as early as 50 years (UNHCR 2025). According to the World Health Organization (WHO), the global population aged 60 years and older is projected to increase from 1 billion in 2020 to 1.4 billion by 2030, and to more than double by 2050, reaching approximately 2.1 billion (WHO 2025). This marked increase in the number of older individuals worldwide highlights the need for a more comprehensive understanding of the ageing process, encompassing physiological, biological, psychological, and social aspects, in order to effectively adapt societies to this demographic shift and create opportunities for promoting healthy ageing.

In line with the scope of this PhD thesis, the following sections will address the general physiological changes that occur in the neuromuscular system during ageing.

1.1.2 Age-related changes in the neuromuscular system

Normal ageing is accompanied by a range of structural and functional changes within the neuromuscular system, including alterations in motoneurons, neuromuscular junctions, and muscle fibres, as shown in **Figure 1** (Hunter et al. 2016). Collectively, these changes have a profound impact on motor performance in older adults (Borzuola et al. 2020).

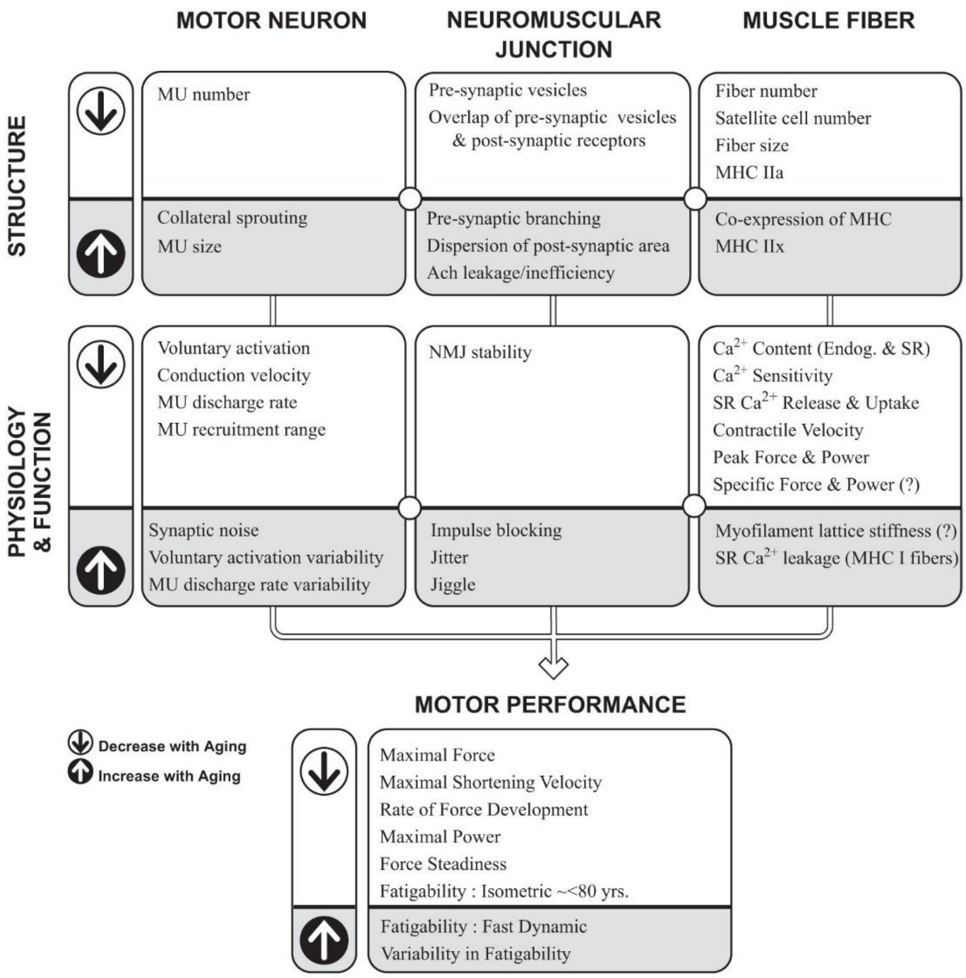

Figure 1. Structural and physiological age-related changes in motor neurons, neuromuscular junctions, and muscle fibres, and their impact on motor performance outcomes.

Source: Hunter et al. (2016).

1.1.2a Changes at motor neuron level

One of the main contributors to sarcopenia, which refers to the progressive and well-documented decline in skeletal muscle mass associated with ageing (Cruz-Jentoft et al. 2019), is the age-related motor unit remodelling resulting from spinal motor neuron apoptosis (Piasecki et al. 2016). Motor neuron apoptosis may begin gradually early in life and tends to accelerate after the age of 60, with more rapid declines in muscle mass and function usually observed between approximately 75 and 80 years of age (Hunter et al. 2016). This neuronal loss leads to denervation of previously innervated muscle fibres, although these fibres are typically reinnervated through collateral axonal sprouting from adjacent surviving motor neurons (Hepple and Rice 2016). The reinnervation process results in an increased muscle fibre density (Stalberg and Thiele 1975) and fibre-type grouping (Lexell and Downham 1991), leading to fewer but larger surviving motor units that contribute to a greater proportion of the overall motor unit pool (McNeil et al. 2005; Barry et al. 2007). However, this compensatory reinnervation process is limited, as it does not allow for the full reinnervation of all muscle fibres that have lost their original motor neuron. For instance, post-mortem studies have shown that approximately 30–40% of fibres in the vastus lateralis are lost by around 75 years of age, with the loss of both type I and type II fibres contributing substantially to muscle atrophy (Lexell et al. 1988). Additionally, morphological changes in the surviving motor units occur due to increased oxidative stress and low-grade inflammation with ageing (Opalach et al. 2010)

Besides motor unit remodelling, ageing is also associated with alterations in the synaptic inputs received by motoneurons. Spinal motor neurons integrate tens of thousands of synaptic inputs originating from descending pathways, spinal interneurons, and sensory afferents, which collectively contribute to the net excitatory drive required for motor output (Hug et al. 2023). Regarding the descending pathways, cortical neuronal atrophy and widespread reductions in both gray and white matter volume have been reported throughout the cortex (Lee and Kim 2022). These structural changes are accompanied by functional alterations, including a significant decline in inhibitory processes and increased activation across multiple cortical areas in older adults when performing a particular task (Grady 2012). The reduced ability to selectively inhibit cortical areas during specific motor tasks, with a consequent more generalised and diffuse cortical activation, can lead to inappropriate motor unit recruitment and altered muscle activation pattens (Seidler et al. 2010). Consistent with this, some previous studies have reported greater muscle activity in older adults compared to younger individuals during various motor tasks of both upper and lower limbs (Shinohara et al. 2003; Marshall et al. 2020; Santos et al. 2021). Therefore, the structural and functional cortical changes that occur with ageing can alter the corticospinal input, ultimately modifying the synaptic inputs received by spinal motor neurons (Hunter et al. 2016). In addition, it has been demonstrated that a portion of these synaptic inputs is shared across multiple motor neurons and is referred to as common synaptic input, whereas the remaining portion, which is not shared, is defined as independent input (Hug et al. 2023). Changes in the common synaptic inputs across motor neurons play a key role in coordinating and regulating muscle activity. Specifically, the common synaptic input drives the discharge rates of motoneurons at a shared low-frequency that influences the force produced by the muscle (Negro and Farina 2011). This common modulation is typically quantified using coherence analysis between two electrophysiological signals in the frequency domain. Several studies have reported an altered ability of older adults to appropriately modulate the common synaptic inputs to spinal motor neurons. For example, Castronovo et al. (2018), using coherence analysis between motor unit spike trains, demonstrated that the reduced force steadiness observed in older adults is associated with age-related increases in the fluctuations of low-frequency common synaptic inputs during submaximal isometric contractions. Indeed, the typical decline in force accuracy observed with ageing appears to be unrelated to motor unit remodelling and is instead primarily attributable to alterations in the synaptic inputs received by motoneurons (Barry et al. 2007). This aspect will be examined in greater detail in Paragraph 1.4, in relation to the specific aims of this thesis.

Regarding reflex responses of spinal motoneurons, afferent feedback from sensory receptors is reduced with ageing, leading to slower and less efficient reflex pathways (Geertsen et al. 2017). For instance, both the amplitude and modulation of Hoffman reflex (H-reflex), which primarily reflects the efficacy of Ia afferents in activating spinal motor neurons, are often reduced in older adults compared to younger individuals (Hunter et al. 2016). In support of this, a recent study of our research group (Scalia et al. 2024) has demonstrated that older adults did not exhibit any modulation of H-reflex responses following an acute session of neuromuscular electrical stimulation, in contrast to the modulation observed in younger controls. This reduced efficacy of Ia afferent transmission to spinal motor neurons in ageing can be accompanied by compensatory mechanisms at cortical level. Indeed, as previously mentioned, greater corticospinal excitability has been reported in older adults compared with young individuals, suggesting an increased reliance on descending cortical drive to control muscle activity (Baudry et al. 2014).

1.1.2b Changes at neuromuscular junction level

The neuromuscular junction (NMJ) is the specialised chemical synapse between an alpha motoneuron and a skeletal muscle fibre. It is well-established that the NMJ undergoes both structural and functional alterations with advancing age (Hepple and Rice 2016). Particularly, ageing is associated with remodelling of the NMJ and alterations in neuromuscular transmission, which may negatively affect motor unit activation (Borzuola et al. 2020). Structural changes include enlarged and more fragmented postsynaptic regions, a reduced density of synaptic vesicles and acetylcholine stores, and impaired alignment between presynaptic vesicle release sites and postsynaptic receptors (Deschenes 2011). Collectively, these alterations lead to a decline in the functional integrity and stability of the NMJ. For example, while the NMJ typically ensures 1:1 transmission of action potentials to the muscle fibre, this transmission reliability seems to decline with ageing (Wood and Slater 2001). In addition, needle electromyographic recordings have been used to assess the stability of neuromuscular transmission in older adults by examining the variability in motor unit potential shape. Specifically, two key parameters have been evaluated: "jitter", which refers to the variability in the timing between action potentials of different muscle fibres within the same motor unit, and "jiggle", which reflects the variability in the overall shape of a motor unit potential across consecutive discharges (Hourigan et al., 2015; Stalberg and Sonoo, 1994). Both parameters have been reported to increase with healthy ageing, suggesting greater NMJ instability and impaired neuromuscular transmission in older adults (Balci et al. 2005; Hourigan et al. 2015). Interestingly, near-fibre jiggle and jitter appear to be more sensitive markers of sarcopenia severity (e.g., distinguishing between pre-sarcopenia and severe sarcopenia) than motor unit number estimates (Gilmore et al. 2017). These alterations are accelerated by menopause and physical inactivity and are detectable before substantial muscle loss and functional decline occur, highlighting their importance for the early identification of age-related neuromuscular decline (Cui et al. 2025).

1.1.2c Changes in muscle fibres

Ageing is associated with a net reduction in innervated muscle fibres and a decrease in the size of fibres within the remaining motor units (Hunter et al. 2016). As previously mentioned, loss of muscle fibres is primarily attributed to the age-related motor unit remodelling, which is characterised by a higher rate of denervation relative to reinnervation (Piasecki et al. 2016). Additionally, the regenerative capacity of muscle fibres declines with ageing due to impaired satellite cell function, which is influenced by both extrinsic and cell-intrinsic alterations (Blau et al. 2015). A reduction in muscle fibre cross-sectional area is also commonly observed with advancing age, primarily due to decreased protein synthesis and a decline in the number of satellite cells (Kadi and Ponsot 2010; Wall et al. 2015). Although this atrophy affects all fibre types, it appears to be more pronounced in type II fibres than in type I fibres (Lexell et al. 1988; Hunter et al. 1999; Verdijk et al. 2007). However, it remains unclear whether this preferential atrophy of type II fibres persists when muscle wasting becomes severe, as the high degree of coexpression of multiple myosin heavy chain isoforms within individual muscle fibres complicates the identification of their original fibre type (Purves-Smith et al. 2014). Moreover, the contractile properties of muscle fibres deteriorate with advancing age. In particular, aged muscle fibres exhibit slower contractile speeds compared to those of younger adults, as evidenced by reduced rates of force development and lower maximal shortening velocity (Larsson et al. 1997; Krivickas et al. 2001; Power et al. 2016). This decline has been typically associated to slower cross-bridge kinetics (Power et al. 2016). Similarly, the rate of muscle relaxation decreases with ageing, primarily due to alterations in cross-bridge mechanics and reduced calcium uptake and calcium-ATPase activity (Hunter et al. 1999). The study by Cogliati et al. (2023) indicated that impairments in muscle relaxation and re-elongation of the tibialis anterior may partly contribute to altered ankle dynamics during transitions between different phases of the gait cycle in older adults.

1.2 Trunk extensor muscles

1.2.1 General overview

The trunk extensor muscles comprise both deep (e.g., multifidus) and more superficial (e.g., erector spinae) muscles located along the posterior part of the trunk (Bogduk 2016). These muscles are fundamental for maintaining spinal integrity and function (Panjabi 1992). Their primary role is to extend the vertebral column, a movement during which the vertebral bodies undergo posterior sagittal rotation accompanied by a small degree of posterior translation (Bogduk and Endres 2005). By generating this extension, trunk extensor muscles counteract the anterior flexion forces exerted by gravity. This function is essential for stabilising the spine and is particularly critical for sustaining an upright posture (Banno et al. 2019; Nakahira et al. 2025). Indeed, Nakahira et al. (2025) demonstrated that trunk extensor muscles play a key role in preventing forward tilting of the head, trunk, and pelvis during upright standing. In addition, these muscles are crucial for the execution of various daily activities, such as lifting objects, as evidenced by several EMG studies (Bonato et al. 2003; Chow et al. 2004; Sanderson et al. 2024). The principal muscle group responsible for trunk extension is the erector spinae (ES), which runs longitudinally along the vertebral column and is composed of three main muscles: the spinalis, longissimus, and iliocostalis (Delp et al. 2001). Each of these components spans multiple spinal regions, from the cervical to the lumbar spine. The multifidus (MFD) muscle is also considered one of the primary muscles involved in spinal extension, although it contributes to this movement to a lesser extent than the ES, as will be discussed in the following section. It extends from the sacrum to the cervical spine, with its thickest portion located in the lumbar region (Hofste et al. 2020).

In accordance with the objectives of the experimental studies presented in this PhD thesis, the following sections will specifically focus on the anatomical and functional characteristics of the lumbar extensor muscles, followed by an in-depth analysis of age-related changes in these muscles.

1.2.2 Lumbar extensor muscles

The lumbar spine is surrounded by several muscles that are typically classified into three groups based on both anatomical positioning and functional roles, highlighting the complexity of this region's musculature (Bogduk and Endres 2005). Specifically, these include: (1) the psoas major, which lies along the anterolateral aspect of the lumbar spine; (2) the intertransversarii laterales and quadratus lumborum, which span and cover the transverse processes anteriorly; and (3) the lumbar back muscles, which are located posteriorly and overlay the vertebral arches and spinous processes. Both the lumbar ES and lumbar MFD muscles belong to the third group. The lumbar ES is composed of two main muscles: the iliocostalis lumborum and the longissimus thoracis (Bogduk and Endres 2005). Each of these muscles can be further subdivided into two anatomical regions based on their origin: a lumbar portion, with fascicles originating from the lumbar vertebrae, and a thoracic portion, with fascicles originating from the thoracic vertebrae or ribs. Accordingly, these four subdivisions are referred to as:

- iliocostalis lumborum *pars lumborum*
- longissimus thoracis *pars lumborum*
- iliocostalis lumborum *pars thoracis*
- longissimus thoracis *pars thoracis.*

In contrast, the MFD muscle is located more medially than the ES and consists of both deep and more superficial muscle fibres (**Figure 2**) (Hodges and Danneels 2019).

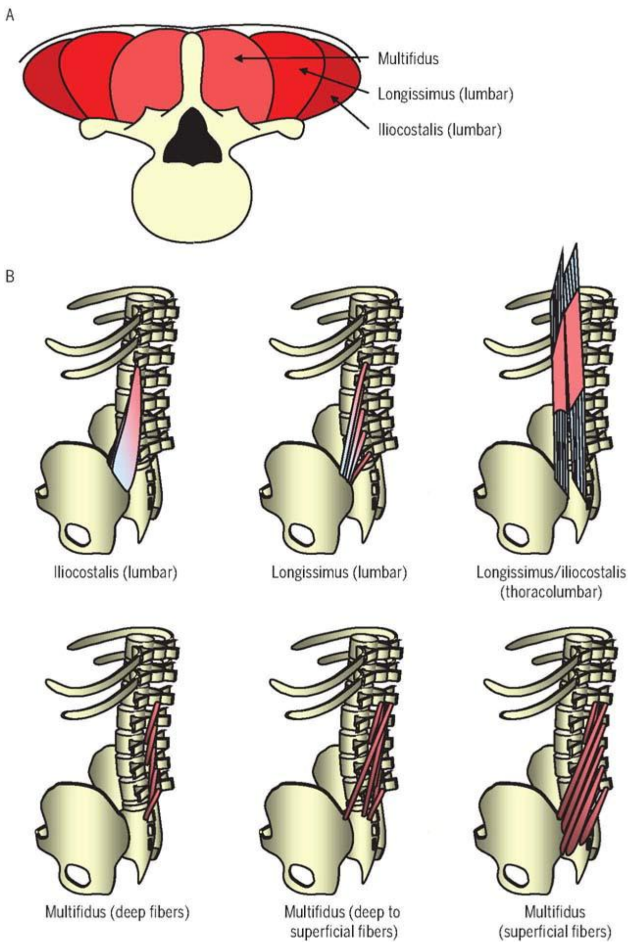

Figure 2. Anatomical description of the lumbar extensor muscles. (A) Cross-sectional view at the L4 lumbar level. (B) In-situ representation showing muscle fibre orientation.

Source: Hodges and Danneels (2019).

Overall, the MFD is characterised by relatively short fibres that are arranged in tightly packed bundles (Hodges and Danneels 2019). This structural configuration results in a physiological cross-sectional area that is twice that of the ES muscle (Mawston and G. Boocock 2015), enabling the MFD to produce high forces over a limited range of motion (Ward et al. 2009). As a result, this muscle is particularly suited for providing intersegmental stability to the spine, rather than generating large-scale spinal extension movements (Ward et al. 2009). Indeed, the MFD muscle has been reported to function primarily as a spinal segmental stabiliser, playing a key role in maintaining static postural alignment, while the ES acts more as a spinal mobiliser, contributing to dynamic control and movement of the spine (Banno et al. 2019).

Although the MFD muscle is critical for lumbar spinal stability, its deep anatomical position limits the ability of using non-invasive assessment techniques such as surface EMG. In contrast, the ES muscle is more accessible to such techniques due to its more superficial location. Therefore, since electromyographic assessments were employed in the experimental studies of this PhD thesis, the emphasis was placed on investigating the activity of the lumbar ES.

1.2.3 Changes in the lumbar extensor muscle morphology with ageing

Several studies have examined the morphological characteristics of the ES and MFD muscles in older adults, focusing on parameters of muscle quantity (e.g., muscle cross-sectional area) and muscle quality (e.g., fat infiltration) (Masaki et al. 2016; Sions et al. 2017; Shahtahmassebi et al. 2017; Han et al. 2024). Overall, findings indicate that lumbar extensor muscles undergo degenerative morphological changes as part of the normal ageing process, primarily characterised by muscle atrophy and increased fat infiltration (Dallaway et al. 2020). Notably, evidence from a systematic review and meta-analysis by Dallaway et al. (2020) suggests that these changes are muscle-specific within the lumbar region, with the ES and quadratus lumborum muscles showing the most pronounced age-related degeneration. However, both the MFD and ES appear to undergo earlier degenerative changes compared to other muscles such as the psoas or quadratus lumborum (Liu et al. 2024). Nevertheless, in accordance with the scope of this thesis, detailed discussion will be limited to the age-related degenerative changes observed in the ES and MFD muscles.

Interestingly, increases in fat infiltration show a stronger correlation with advancing age than reductions in muscle size within the lumbar muscles, making fat infiltration a more prominent degenerative change associated with ageing in this region (Dallaway et al. 2020). This highlights fat infiltration as a potentially more sensitive biomarker of age-related degeneration in the paraspinal musculature. In line with this, previous MRI-based research has shown that paraspinal muscles exhibit a significantly greater increase in fat infiltration with ageing compared to leg muscles, suggesting a higher susceptibility to age-related degenerative changes (Dahlqvist et al. 2017). Conversely, lower limb muscles (e.g., quadriceps, hamstrings, ankle extensors) tend to undergo more pronounced atrophy than the lumbar back muscles (Leblanc et al. 1992; Abe et al. 2014). One possible explanation for this phenomenon is the high proportion of type I (slow-twitch) muscle fibres in the lumbar extensor muscles (approximately 60–70%) (Jørgensen et al. 1993), which may influence their susceptibility to fat infiltration while offering greater resistance to atrophy. Indeed, slow-twitch muscle fibres tend to accumulate a greater amount of intramyocellular lipid droplets with advancing age compared to fast-twitch fibres (Gueugneau et al. 2015; Choi et al. 2016), whereas the latter typically exhibit greater atrophy (Lexell et al. 1988; Gueugneau et al. 2015). In addition, the degree of fat infiltration in the lumbar paravertebral muscles is not uniform but varies according to the specific spinal level at which it is assessed

(Dallaway et al. 2020). Specifically, measurements taken at the upper lumbar levels (L1–L2) tend to show less age-related fat infiltration, whereas the mid (L2/3–L3/4) and lower (L4–L5/S1) lumbar levels exhibit greater increases in fat infiltration with advancing age. This suggests that the mid-to-lower lumbar regions may be more susceptible to age-related degenerative changes. These findings have important implications for both research and clinical assessments, as evaluating only the upper lumbar levels may underestimate the extent of degeneration, whereas including the mid and lower lumbar levels provides a more accurate and comprehensive representation of age-related changes in muscle quality.

1.2.3a Functional implications

The age-related changes in lumbar muscle morphology described in the previous section have important implications not only for spinal function but also for overall physical performance. A cross-sectional study by Hicks et al. (2005a) demonstrated that high levels of fat infiltration in trunk muscles - including the paraspinal, rectus abdominis, and lateral abdominal muscles - were significantly associated with poorer physical function in healthy older adults, whereas reductions in muscle mass were not. This association was observed in both objective measures (Health ABC Physical Performance Battery) and self-reported assessments of functional ability. Notably, fat infiltration in the trunk muscles accounted for approximately 13% of the variance in physical performance, compared to only 6% explained by fat accumulation in the thigh muscles. This suggests that trunk muscle composition may have a greater impact on the performance of tasks typically attributed to lower extremity function. In addition, poor trunk muscle quality was also associated with a history of low back pain (LBP) among participants. Taking a step further, a longitudinal study by the same authors demonstrated that older adults with higher levels of fat infiltration in the trunk muscles exhibited reduced functional capacity, particularly in balance, three years later (Hicks et al. 2005b). This association was especially pronounced in individuals with a history of LBP and was independent of thigh muscle composition. In relation to LBP, previous research has also shown that older adults with chronic LBP exhibit greater intramuscular fat in the MFD, and reduced size of the ES compared to age-matched individuals without LBP (Sions et al. 2017). In addition, Masaki et al. (2016) found that maximal walking speed in middle-aged and older women was significantly associated with the thickness of lumbar ES muscle. More recent studies have also demonstrated a significant negative correlation between fat infiltration in the paraspinal muscles (lumbar MFD and ES) and trunk extensor endurance both in individuals without LBP across various age groups (Han et al. 2024) and in individuals with low back problems (Yazici and Yerlikaya 2022). Notably, Yazici and Yerlikaya (2022) reported that individuals with less than 10% fat infiltration in the paraspinal muscles exhibited the longest endurance times, whereas those with greater than 50% fat infiltration showed the shortest durations.

1.3 Muscle fatigue

1.3.1 General overview and physiological mechanisms

Fatigue is a multifaceted concept that includes responses to both physical and psychological workloads, and it generally represents a biological reaction to intensive and/or prolonged activity (Avlund 2013). Although the concept of fatigue has acquired a multitude of definitions in the literature depending on the field of research, a detailed analysis of its different meanings is beyond the scope of this PhD thesis. Instead, the focus here will be primarily on muscle fatigue, defined as “any exercise-induced reduction in the ability of a muscle to generate force or power” (Gandevia 2001). Muscle fatigue is a complex phenomenon that results from alterations occurring at different levels within the neuromuscular system. These alterations affect both central physiological processes involved in muscle activation (e.g., excitability of the motor cortex and spinal α-motoneurons, descending corticospinal pathways, afferent feedback) and peripheral processes contributing to contractile function (e.g., sarcolemmal excitability, mechanisms related to the cross-bridge cycle, muscle metabolism) (Gandevia 2001). In particular, the site of impairment depends on the nature of the task performed. According to the task dependency principle, there is no single universal cause of muscle fatigue; rather, the dominant mechanism is determined by the physiological processes undergoing the highest amount of stress during the specific fatiguing task (Enoka and Duchateau 2008). Muscle fatigue is viewed both as a "negative" consequence of physical efforts and as a "positive" protective mechanism that helps prevent injury or adverse outcomes during exercise (Avlund 2013). In general, the ability to resist muscle fatigue during sustained or repetitive tasks is essential for a wide range of activities, from daily living to athletic performance. Consequently, the investigation of fatigue development and its underlying mechanisms has been a topic of great interest in research over time.

In line with the scope of this PhD thesis, the following sections will first examine the general age-related differences in muscle fatigue, followed by a more detailed exploration of age-related changes in fatigue and endurance of the trunk extensor muscles.

1.3.2 Age-related changes in muscle fatigue

Age-related differences in the development of muscle fatigue have been widely investigated given the important role muscle fatigue plays in the functional independence of older adults (Theou et al. 2008). As described in previous sections, ageing is associated with changes in motor unit inputs, morphology, and physiology, which can alter the rate at which motor units are stressed during a fatiguing task (Hunter et al. 2016). Notably, despite the typical decline in muscle mass and strength that occur with ageing, several studies showed greater fatigue resistance in older people compared to younger adults during maximal and submaximal isometric contractions of both upper and lower limb muscles (Christie et al. 2011). For example, healthy older men and women exhibited less fatigue than their younger counterparts during intermittent maximal voluntary isometric contractions (Callahan et al. 2009) and submaximal sustained contraction (Mcphee et al.

2014) of the knee extensors. Similarly, older adults demonstrated longer endurance times than younger adults during both maximal isometric contractions (Bilodeau et al. 2001) and submaximal contractions (Bazzucchi et al. 2005) of the elbow flexors. This increased fatigue resistance, referred to as the "fatigue paradox", has been attributed to age-related physiological changes, such as greater loss and atrophy of fast-twitch muscle fibres, motor unit remodelling, and a reduced reliance on glycolytic metabolism (Hunter et al. 2016). All these factors contribute to slower contractile properties in older adults and to a lower accumulation of inorganic phosphate and hydrogen ions during contraction, which can interfere with force production (Lanza et al. 2007). In addition, it has been suggested that the greater fatigue resistance observed in older adults may result from their lower absolute muscle forces compared to younger individuals (Kent-Braun 2009). During standardised exercise protocols (e.g., at a fixed percentage of maximal voluntary contraction), stronger individuals are required to sustain higher relative workloads than their weaker counterparts. This greater mechanical demand may increase the intramuscular pressure, thereby restricting perfusion and oxygen delivery, and promoting the accumulation of metabolites (Kent-Braun et al. 2002). Nevertheless, some previous studies have demonstrated that the age-related changes in muscle fatigue development cannot be attributed to reductions in muscle strength (Bautmans and Mets 2005; Hunter et al. 2005; Katsiaras et al. 2005; Chung et al. 2007). For example, Chung et al. (2007) reported that even when subgroups of young and older men were matched for baseline strength, older adults still exhibited greater fatigue resistance during maximal intermittent isometric contractions of the ankle dorsiflexors.

However, the effect of ageing on muscle fatigue depends on the type of contraction performed. Indeed, the increased fatigue resistance observed during isometric contractions with advanced age does not appear to extend to dynamic contractions (Christie et al. 2011). Understanding age-related differences in fatigue development during dynamic contractions is more complex than isometric tasks, as it is influenced by a greater number of task parameters such as range of motion and angular velocity. Interestingly, a review by Paris et al. (2022) reported consistently greater fatigue in older adults compared with younger controls during isotonic contractions, whereas tasks that constrain angular velocity (isokinetic contractions) did not yield the same consistent results. This suggests that angular velocity is a particularly important factor influencing age-related differences in muscle fatigue during dynamic contractions. However, the specific central (e.g., voluntary activation, motor unit properties) and peripheral (e.g., metabolite accumulation, sarcolemmal membrane excitability) mechanisms underlying age-related differences in muscle fatigue during dynamic contractions remain poorly understood (Paris et al. 2022).

1.3.3 Age-related changes in muscle fatigue of the trunk extensor muscles

Although the previous sections (1.2.2 and 1.2.3) focused on lumbar extensor muscles, this section considers the trunk extensor muscles more generally, as endurance tests engage both the lumbar and thoracic musculature rather than isolating a single spinal segment. The importance of trunk extensor endurance capacity has been highlighted in previous studies (Das 2016; Ghamkhar and Kahlaee 2019), given that these muscles are typically exposed to sustained or repetitive low-level contractions to maintain upright posture (Falla et al. 2014). For instance, fatigue of the trunk extensor muscles has been reported to impair postural control during both static (Lin et al. 2009; Johanson et al. 2011) and dynamic tasks (Granata and Gottipati 2008). In addition, reduced endurance of the back muscles has been shown to predict long-term back-related disability (Enthoven et al. 2003).

Given the critical role of the trunk extensor muscles in preserving spinal health, their function has also been evaluated in the context of ageing. Notably, Beauchamp et al. (2016) identified endurance of the trunk extensor muscles as a key physical factor influencing older adults' participation in social life. Furthermore, trunk extensor endurance has been reported to be a predictor of balance performance (Flora et al. 2022) and a determinant of functional ability in older adults (Suri et al. 2011; Mesquita et al. 2019).

Several studies have investigated the age-related changes in trunk extensor muscle endurance. In this context, a systematic review and meta-analysis was conducted and will be presented in Chapter 3. Unlike the experimental studies of this PhD thesis, which focused on the trunk (and specifically the lumbar region for the EMG assessments), the review considered the spinal extensor muscles more extensively, including also the cervical region, providing a more comprehensive overview of the entire extensor musculature of the spine.

1.4 Muscle force control

1.4.1 General overview and physiological mechanisms

Muscle force control refers to the ability of the neuromuscular system to produce, modulate, and maintain a desired level of force or torque during voluntary muscle contractions (Enoka and Farina 2021). However, even when a person attempts to exert a constant force, the output is not perfectly steady but fluctuates around the intended target (Enoka and Farina 2021). These fluctuations are typically quantified during submaximal contractions against resistance, using magnitude-based measures, such as the standard deviation and coefficient of variation, which provide indices of force or torque steadiness. Such fluctuations have important implications for task performance across a range of contexts and force levels (Pethick et al. 2022).

Skeletal muscles protect bones, cartilage, and joints by absorbing mechanical impacts and resisting excessive tension forces (McQuade and Murthi 2004; Rudenko et al. 2016). Optimal neuromuscular control and force generation are therefore essential for maintaining joint integrity and preventing injury during a given physical task (Clark et al. 2023). Consequently, these abilities are crucial for preserving overall physical function, and their deterioration can compromise the precision of voluntary movements, negatively affecting joint stability, coordination, and motor performance (Arvanitidis et al. 2024). For this reason, muscle force control has been widely investigated across different populations, including older adults and individuals with musculoskeletal conditions.

From a physiological perspective, muscle force generation is regulated by the central nervous system through size-ordered motor unit recruitment (Henneman et al. 1965) and modulation of motor neuron discharge rates (Person and Kudina 1972; Milner-Brown et al. 1973). Interestingly, only the low-frequency components of the neural drive to the muscle are reflected in the motor output (Mannard and Stein 1973). Common low-frequency components in motor neuron discharge rates have been observed and termed *common drive* (De Luca et al. 1982). De Luca and Erim (1994) proposed the concept of common drive as a possible motor unit control strategy, postulating that the distinct firing patterns of individual motor units are not governed by separate command signals, but instead arise from a shared common drive to which motor units respond differently. Building on this concept, Negro et al. (2009) and Farina et al. (2014) showed that the effective neural drive to the muscle reflects the common synaptic input to motor neurons. They further demonstrated that this widely distributed common input represents the primary determinant of muscle force control (Farina and Negro 2015). Considering all of the above, the low-frequency components (0-10 Hz) of the neural drive to the muscle are primarily responsible for force generation, as they reflect the common synaptic input received by the motoneuron pool. This common input determines the precise neural command required for optimal force generation, while oscillations within this input challenge the control of the force output (Farina and Negro 2015). Consequently, force variability is attributed to fluctuations in the low-frequency components of the neural drive (Farina and Negro 2015). These physiological mechanisms will be revisited in Section 1.5.3c, where coherence analysis is discussed, particularly in the context of trunk muscles.

In accordance with the scope of this PhD thesis, the following sections will first address the general age-related differences in muscle force control before examining in greater detail the changes in force control of the trunk extensor muscles.

1.4.2 Age-related changes in muscle force control

Older adults typically exhibit lower force steadiness than younger individuals (Pethick et al. 2022). Several studies have reported greater force fluctuations in older adults during sustained submaximal isometric contractions of both upper- and lower- limb muscles (Hunter et al. 2016). For example, reduced force steadiness has been reported in older adults during low-force isometric contractions of the knee extensor muscles (Tracy and Enoka 2002). Similarly, older individuals demonstrated a reduced ability to maintain constant submaximal forces with the first dorsal interosseous muscle (Galganski et al. 1993; Laidlaw et al. 2000). Notably, previous research has shown that the age-related increase in force fluctuations is influenced by contraction intensity (Pethick et al. 2022). Specifically, these fluctuations appear to be most pronounced at lower contraction intensities, with large effects becoming evident up to approximately 35% MVC. At higher contractions intensities (up to 80% MVC), the effects are generally smaller and less consistent (Oomen and van Dieën 2017). This age-related increase in force fluctuations (i.e., less force steadiness) at lower contraction intensities is particularly relevant, as most activities of daily living, especially those commonly performed by older adults, typically require forces of up to approximately 20% MVC (Pethick et al. 2022). Furthermore, age-related reductions in force steadiness are further amplified when older adults perform dual tasks that combine a force maintenance task with a cognitive challenge (Vanden Noven et al. 2014; Pereira et al. 2015). These findings are particularly relevant for daily and occupational activities that require concurrent cognitive demand and submaximal isometric efforts (Vanden Noven et al. 2014).

Although isometric contractions have been the most commonly used paradigm to investigate force control in older adults, dynamic contractions have also been studied. Interestingly, the increase in force fluctuations with ageing is typically greater during sinusoidal, concentric and eccentric tasks than during isometric contractions (Pethick et al. 2022). For example, the steadiness during slow dynamic contractions was lower in older adults compared with younger adults in the first dorsal interosseous muscle, particularly at the lightest loads (Laidlaw et al. 2000). Additionally, steadiness during lengthening contractions was reduced compared to shortening contractions in older adults, whereas no such difference was observed in younger adults (Laidlaw et al. 2000). Similarly, Hortobágyi et al. (2001) reported that older adults exhibited substantially impaired force accuracy compared with younger adults during low-level isokinetic quadriceps contractions. Moreover, older adults showed nearly twice the error during concentric contractions and nearly three times the error during eccentric contractions compared with younger adults (concentric: older 9 ± 3 N, young 5 ± 2 N; eccentric: older 29 ± 14 N, young 10 ± 4 N). This difference in force control between concentric and eccentric contractions with ageing may have important functional implications for activities of daily living, such as descending stairs in the case of lower-limb muscles (Pethick et al. 2022).

From a physiological perspective, as mentioned in the previous section, variability in the common synaptic input to motor neurons is the main determinant of force fluctuations (Farina and Negro 2015). Accordingly, Castronovo et al. (2018), using high-density electromyography decomposition analysis, indicated that reduced force steadiness in older individuals during submaximal contractions of the dorsiflexor muscles was associated with increased fluctuations in the common synaptic input to motoneurons. Similarly, Feeney et al. (2018) found that longer pegboard times in older adults were associated with reduced force steadiness and greater fluctuations in the estimated common synaptic input to motor neurons during steady contractions. These findings suggest that alterations in the common synaptic input can explain a large part of the impaired force control observed in older adults (Pethick et al. 2022).

1.4.3 Age-related changes in force control of the trunk extensor muscles

The trunk extensor muscles counteract the anterior flexion forces imposed by gravity, which is essential for stabilising the spine and maintaining an upright posture (Banno et al. 2019; Nakahira et al. 2025). Impaired spinal sensorimotor control may lead to abnormal tissue loading and increased mechanical strain, potentially resulting in pain or injury (Arvanitidis et al. 2022). Consequently, optimal activation of the trunk extensor muscles is critical for maintaining smooth force output and minimising mechanical overload of the lumbar spine.

In the context of ageing, previous research has shown that older adults exhibit greater force fluctuations of the trunk extensor muscles compared with younger individuals. In particular, this age-related decrease in force steadiness has been demonstrated during a fatiguing task at 30% of maximal voluntary isometric force (Parrella et al. 2025) and during a 15-s steadiness task at 10% of peak torque (Porto et al. 2020). Additionally, trunk position and force sense have been shown to be altered in older individuals with hyperkyphosis (Keshavarzi et al. 2022). Interestingly, Forestieri Faccio et al. (2021) reported that better torque steadiness at 10% of peak torque was associated with improved posterior postural stability in older adults. The same authors also observed that torque steadiness was overall greater in the trunk flexor muscles than in the trunk extensors, suggesting that force control of the extensor muscles may be more affected by age. Moreover, Cangussu-Oliveira et al. (2020) reported that among the trunk muscle parameters, torque steadiness of the extensors had the strongest influence on the presence of vertebral fractures. Lastly, the ability of the trunk to control submaximal isometric torque was positively associated with the performance of various functional tasks in community-dwelling older women, including forward and lateral stepping and the 5-times stand-to-sit test (de Abreu et al. 2025).

However, all these studies have focused on isometric trunk extension contractions, while it remains unclear whether these findings can be extended to dynamic trunk extension movements.

1.5 EMG technique

1.5.1 General overview

EMG is an electrophysiological technique used to record and assess the electrical activity of skeletal muscle cells during contraction by detecting changes in electric potentials (Nazmi et al. 2016). This technique enables the investigation of neuromuscular activity and provides insight into muscle responses to neural stimuli under various conditions (Reaz et al. 2006). EMG recordings are complex signals that depend on the integrity of the nervous system and are influenced by the anatomical and physiological properties of the muscles (Chowdhury et al. 2013).

There are two modalities to acquire myoelectric signals: needle EMG and surface EMG (sEMG). The former is an invasive technique, whereas the latter is a non-invasive technique. In line with the experimental procedures of this PhD thesis, the following sections will focus on sEMG only, with particular emphasis on high-density EMG.

1.5.2 sEMG technique

sEMG is widely used in biomedical settings and research due to its non-invasive and painless nature (Ahmad and Chappell 2009). It allows the generation of a two-dimensional representation of electrical activity, which is sampled both in space by the electrodes and time by an electronic sampler, producing an “electrical image” that evolves over time (Merletti and Muceli 2019). The electrodes are placed directly on the skin, providing information about superficial muscle activation (Chowdhury et al. 2013). More specifically, sEMG signals reflect the summation of all motor unit action potentials (MUAPs) from the muscle beneath the skin (Reaz et al. 2006). The frequency spectrum of sEMG signals typically ranges from 0 to 400 Hz, depending on factors such as electrode spacing, muscle fibre-type composition, subcutaneous fat thickness, and the shape of action potentials (Turker 1993; De Luca et al. 2010).

Traditionally, the most commonly used sEMG configuration is the bipolar one, in which two large electrodes are placed on the skin over the contracting muscle, aligned parallel to the muscle fibre direction (Gerdle et al. 1999). The typical inter-electrode distance ranges from 20 to 30 mm. However, this configuration has limited selectivity, as the relatively large detection volume of each electrode reduces the ability to distinguish activation patterns within specific muscle regions, such as across different spinal levels of the ES muscle, or among adjacent muscles (Falla and Gallina 2020). In addition, measurement repeatability may be compromised when using this configuration due to potential variability in inter-electrode distance (Besomi et al. 2019). To address these limitations, advances in technology and methodology have led to the development of new techniques that improve measurement accuracy and deepen our understanding of neuromuscular activity.

1.5.3 High-density EMG (HDsEMG) technique

HDsEMG involves the simultaneous recording of at least four surface electromyographic signals using small-diameter electrodes (0.5-3 mm) arranged in a bidimensional grid with close inter-electrode spacing (normally 2.5-10 mm) (**Figure 3**) (Gallina et al. 2022).

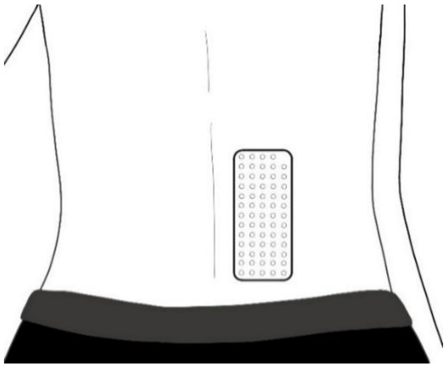

Figure 3. High-density electromyography grid of 64 electrodes placed over the lumbar erector spinae muscle.

Source: Parrella et al. (2025).

This configuration enhances the selectivity of the recordings and allows the assessment of muscle activity over a larger area compared to traditional bipolar sEMG (Falla and Gallina 2020). Due to its high spatial sampling resolution, HDsEMG can be used to characterise several features of the neuromuscular system, including regional activation, muscle fibre properties, and single motor unit discharge characteristics (Gallina et al. 2022). Therefore, HDsEMG represents an ideal technique to investigate the distribution of activity within superficial muscle groups, such as the ES muscle, and to examine how this distribution is altered under different conditions, including fatigue or pain (Falla and Gallina 2020). Moreover, HDsEMG allows a deeper understanding of the physiological mechanisms underlying alterations in muscle force control across various contexts, including ageing. Indeed, as discussed in section 1.4.1, such alterations are associated with changes in motor unit behaviour, particularly in recruitment strategies and discharge rates (Enoka and Duchateau 2017).

In line with the scope of this PhD thesis, the following sections will first address regional activation analysis, which was the main analytical approach applied in the first experimental study (Chapter 2), specifically in relation to muscle fatigue and ageing. Secondly, coherence analysis will be discussed, as this method was employed to investigate age-related alterations in muscle force control in Chapter 4.

1.5.3a Regional activation of the muscle

Regional activation refers to the recruitment and modulation of motor units within a specific region of a muscle or across adjacent muscles (Gallina et al. 2022). This phenomenon occurs because most of the muscle fibres innervated by a single motor neuron are clustered within a particular area of the muscle, rather than being evenly distributed across its volume (Falla and Gallina 2020). For example, Abboud et al. (2020) have demonstrated the presence of cranial and caudal motor unit territories within the longissimus pars lumborum muscle, which can be selectively activated by the central nervous system during voluntary movements. In addition, factors such as fibre type distribution, sarcomere length, muscle architecture (e.g., fascicle length) and tendon-aponeurosis properties further contribute to the regional heterogeneity of contractile function (Sahinis et al. 2025). A summary of these factors is presented in **Figure 4**.

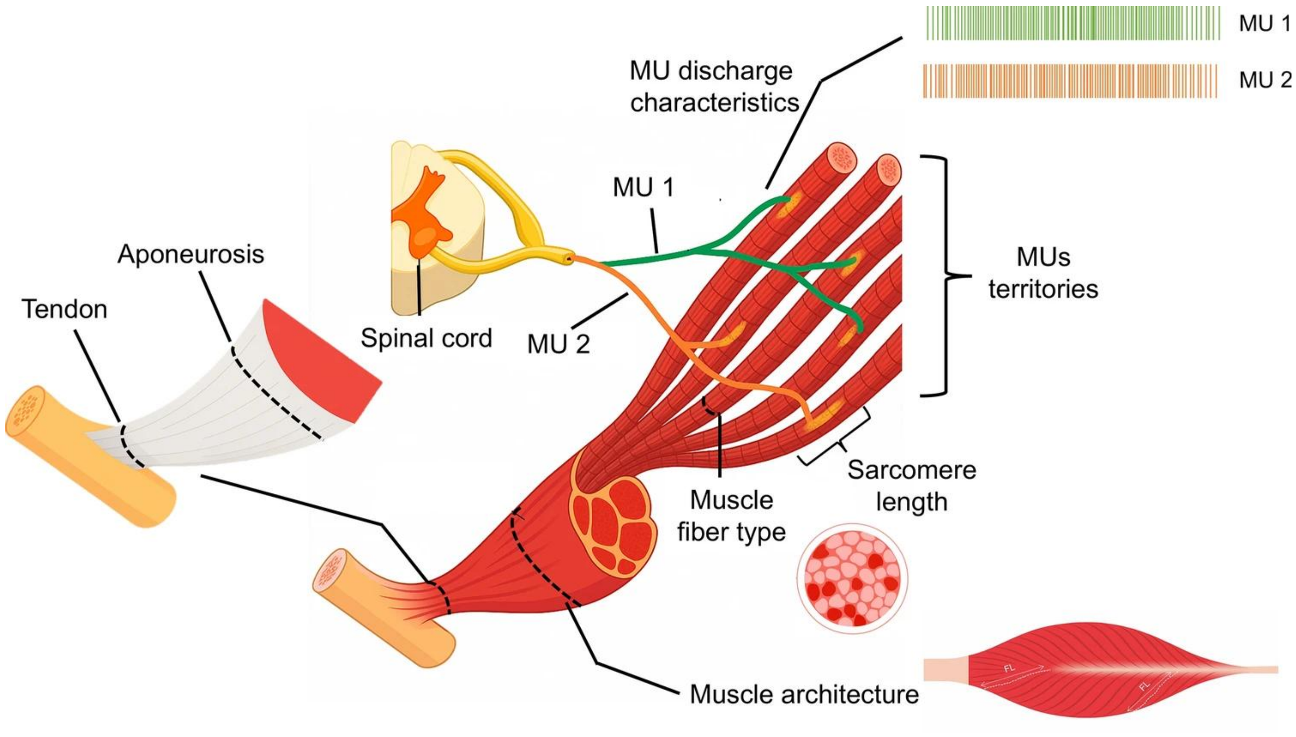

Figure 4. Illustration of the key neuromuscular factors, including compartmentalised neural control, muscle fibre type distribution, sarcomere length, muscle architecture and tendon–aponeurosis properties, that contribute to region-specific differences within a muscle.

Source: Sahinis et al. (2025).

Regional recruitment of muscle fibres can be identified in HDsEMG signals as an amplitude distribution localised above the active fibres (Vieira et al. 2011; Rodriguez-Falces et al. 2013). Accordingly, variations in EMG amplitude can be interpreted as reflecting changes in the activation of muscle fibres across different regions of the muscle (Holtermann et al. 2005; Madeleine et al. 2006). The use of HDsEMG to investigate regional activation within a muscle is supported by findings showing that shifts in EMG amplitude distribution toward a specific region are associated with an increased number of motor units recruited in that area (Falla and Farina 2008) and/or decreased discharge rates of motor units in other regions (Dideriksen et al. 2016).

Common parameters extracted from HDsEMG to assess regional activation include the location, size, and amplitude of active regions (Gallina et al. 2022). More specifically, HDsEMG allows the creation of a topographical representation of muscle electrical activity that shows the relative changes in the intensity of activity within a muscle or muscle group (Falla and Gallina 2020). Thus, a topographical map of HDsEMG amplitude can be generated to characterise the spatial distribution of muscle activity, from which the centroid and/or entropy of the signals are calculated. These parameters will be discussed in the next section.

1.5.3b Topographical maps of HDsEMG activity

A topographical map of HDsEMG activity is generated by mapping the root mean square (RMS) values of the signals, which indicate the magnitude of muscle activation (Farina et al. 2004). From this map, the centroid is typically extracted, representing the weighted central point of muscle activity based on the RMS distribution (Falla and Gallina 2020), as illustrated in **Figure 4**. Thus, an increase in EMG amplitude within a specific region of the electrode grid results in a displacement of the centroid toward that area. The position of the centroid is expressed in a Cartesian coordinate system, with its x- and y-coordinates describing shifts along the medial-lateral and cranial-caudal axes, respectively. These centroid displacements are typically quantified in millimetres. Importantly, even displacements of a few millimetres can reflect considerable changes in EMG amplitude distribution, indicating substantial alterations in muscle activation along and/or across muscle fibres (Falla and Gallina 2020).

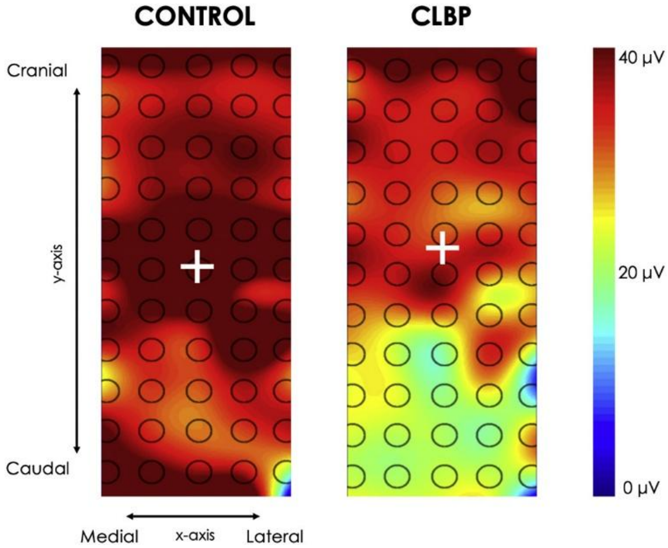

Figure 4. Examples of topographical maps of erector spinae muscle activity obtained from a 13 × 5 electrode grid in a healthy control group (left) and in individuals with chronic low back pain (right) during an isokinetic fatiguing task. The white cross indicates the centroid of muscle activity, aligned with the corresponding coordinate system (x- and y-axes).

Source: Arvanitidis et al. (2021).

RMS values can be further processed to compute the entropy of the HDsEMG signals, providing additional information regarding the spatial distribution of muscle activity. Specifically, entropy reflects the degree of homogeneity in muscle activity, with higher values indicating a more uniform distribution of activity across the electrode grid (Farina et al. 2008).

Topographical maps of HDsEMG activity have been extensively investigated in various contexts, including musculoskeletal pain and muscle fatigue (Gallina et al. 2022). In relation to muscle fatigue, which is of greater relevance to the present thesis, variations in the spatial distribution of muscle activity have been demonstrated across different muscles and experimental conditions, such as during sustained fatiguing contractions or contractions performed at progressively increasing loads (Farina et al. 2008). These findings suggest heterogeneity in the distribution of motor units and/or adaptations in motor control strategies, including inhomogeneous motor unit recruitment or substitution. Notably, Tucker et al. (2009) reported that the activity of the lumbar

ES muscle exhibited spatial adaptations during sustained fatiguing contractions, whereas such adaptations were not observed during contractions performed at varying loads. From a functional perspective, this redistribution of muscle activity may represent a mechanism to maintain the target force and prolong the contraction despite fatigue development, thereby preventing overload of specific muscle regions (Falla et al. 2014). Indeed, Farina et al. (2008) found a positive correlation between the amount of change in the spatial distribution of upper trapezius muscle activity and endurance time during an isometric sustained contraction.

1.5.3c Coherence analysis

In many research contexts, understanding the relationship between two signals is of primary interest. Such relationships can be examined using time-domain approaches, such as cross-correlation, or frequency-domain approaches, such as coherence analysis (El-Gohary et al. 2006).

EMG signals are composed of different frequency-specific components that collectively produce complex waveform patterns (Reaz et al. 2006). By applying frequency analysis (i.e., spectral analysis), which converts the raw EMG signal into its frequency spectrum, the relative contribution of each frequency component can be quantified. Within the total EMG frequency spectrum (up to 500 Hz) (Muceli and Merletti 2024), four main frequency bands have been identified: delta (1-5 Hz), alpha (8-12 Hz), beta (13-21 Hz), and gamma (32-100 Hz) (Castronovo et al. 2018). Importantly, each of these bands reflects distinct underlying physiological processes. The delta band, which is the one considered in Chapter 4 of this thesis, is the most relevant for force generation (Farina and Negro 2015). The alpha band is thought to reflect afferent feedback (Vallbo and Wessberg 1993; Williams and Baker 2009). The beta band likely reflects oscillatory activity within the corticospinal pathways (Farmer et al. 1993; Salenius et al. 1997). Lastly, the gamma band has been associated with neural activity in the primary motor, supplementary motor, and premotor cortices (Feige et al. 2000). Therefore, decomposing the EMG signal into frequency bands allows for a more detailed physiological interpretation.

Taking this a step further, coherence analysis extends traditional spectral analysis by quantifying the frequency-dependent linear relationship between two signals. Specifically, the coherence function, commonly expressed as the magnitude-squared coherence (MSC), describes the degree to which two signals are linearly correlated at each frequency component (Malekpour et al. 2018). The MSC values range from 0, indicating no correlation between the signals, to 1, indicating perfect correlation. This kind of analysis has various applications, as it allows researchers, for instance, to examine the synchronisation and shared neural activity between different muscles, as well as between muscle activity and the generated force.

In the context of muscle force control, coherence analysis is typically employed to investigate mechanisms underlying reduced force steadiness in different contexts (e.g., ageing) by quantifying the correlation between motor unit discharge times and the generated force in the relevant frequency band (Castronovo et al. 2018). However, this approach requires the use of specific decomposition algorithms to accurately identify individual motor unit discharges from the HDsEMG signal (Del Vecchio et al. 2020). Unfortunately, decomposing HDsEMG signals from the ES muscles remains difficult, likely due to the complex spinal anatomy (e.g., the presence of multiple muscle layers and thoracolumbar fascia) and the volume conductor characteristics of the lower lumbar region (e.g., thick subcutaneous layer) (Christophy et al. 2012). These factors can reduce the distinctiveness of motor unit action potential waveforms, thereby limiting the accuracy and applicability of decomposition algorithms (Del Vecchio et al. 2020).

To overcome this limitation, examining the relationship between the overall sEMG signal (rather than individual motor unit discharge times) and force in the frequency domain represents a valuable alternative for understanding the interaction between muscle activity and force production. Specifically, previous studies have demonstrated that the low-frequency force fluctuations are correlated with the low-frequency components of the rectified interference sEMG (Yoshitake and Shinohara 2013; Moon et al. 2014). Therefore, this type of analysis was employed in the experimental study presented in Chapter 4 of this thesis to indirectly estimate the strength of common rhythmic synaptic inputs to the motor unit pool in the ES muscle and to examine their relationship with force production (Arvanitidis et al. 2022). In particular, as previously mentioned, the analysis focused on the delta frequency band (0-5 Hz). Additionally, the use of HDsEMG provided a higher spatial sampling resolution compared with traditional sEMG, which has been shown to enhance the accuracy of sEMG-based force estimation (Staudenmann et al. 2006).

1.6 Rationale and specific aims of the experimental chapters

The proper functioning of the trunk muscles is fundamental to maintain overall functional capacity, as these muscles provide the proximal stability required for effective distal mobility (Forestieri Faccio et al. 2021). In particular, the trunk extensor muscles play a crucial role in counteracting the anterior flexion moments imposed by gravity, thereby allowing the maintenance of an upright posture and contributing to spinal stability (Banno et al. 2019; Nakahira et al. 2025). Despite their critical importance, the neuromuscular function of the trunk extensor muscles has been predominantly examined in clinical populations, such as individuals with chronic low back pain, whereas comparatively less attention has been given to understanding their specific adaptations in the context of ageing. Therefore, the overall aim of this PhD thesis was to enhance the understanding of how endurance capacity and force control (torque steadiness) of the trunk extensor muscles change with ageing, and to investigate the potential neuromuscular mechanisms underlying these changes using HDsEMG.

The studies presented in Chapters 2, 3, and 4 address the following research questions:

Chapter 2

Rationale. Endurance of the trunk extensor muscles is a key determinant of functional ability in older adults (Suri et al. 2011; Mesquita et al. 2019). However, the neuromuscular mechanisms underlying fatigue resistance in these muscles with ageing remain limited, as previous studies did not always include EMG assessments (Champagne et al. 2009; Parreira et al. 2013, 2014) or relied on traditional bipolar sEMG (Singh et al. 2011; Tsuboi et al. 2013; Habenicht et al. 2020), which provides limited information from a small muscle region. In contrast, HDsEMG offers a more comprehensive understanding of muscle responses to fatigue by enabling the analysis of the spatial distribution of muscle activity within a muscle or muscle group (Cè et al. 2020). Redistribution of muscle activity is considered an important adaptive mechanism to sustain force output during fatigue (Farina et al. 2008). To the best of the authors' knowledge, no previous studies have examined the spatial adaptations of trunk extensor (i.e., lumbar erector spinae) activity during fatiguing contractions in older adults.

Research question. Does spatial redistribution of lumbar erector spinae activity differ between older and younger adults during a fatiguing submaximal isometric trunk extension contraction?

Chapter 3

Rationale. The endurance capacity of the spinal extensor muscles is crucial for spinal health and function, as these muscles are continuously exposed to repeated or prolonged activation due to their critical role in maintaining an upright posture (Das 2016; Ghamkhar and Kahlaee 2019). Given their functional importance, several studies have investigated age-related changes in spinal extensor muscle endurance. However, findings across studies remain mixed and inconsistent.

Research question. Do the development of fatigue and endurance differ between older and younger adults in the spinal extensor muscles?

Chapter 4

Rationale. The trunk extensor muscles are the primary supportive muscles of the spine, as they are essential for maintaining spinal stability and posture (Banno et al. 2019;

Nakahira et al. 2025). Previous research has shown that older adults exhibit greater force fluctuations in these muscles than younger individuals (Porto et al. 2020; Parrella et al. 2025). However, studies on trunk extensor force or torque steadiness during ageing are limited and have focused exclusively on isometric contractions, whereas dynamic contractions more closely reflect the functional demands of daily activities. Moreover, the underlying neuromuscular mechanisms remain largely unexplored. In this context, assessing HDsEMG-torque coherence offers a useful approach to better understand the interaction between muscle activity and force production (Arvanitidis et al. 2022).

Research questions. Does force steadiness during isometric and isokinetic submaximal trunk-extension contractions at different intensities differ between older and younger adults? Does HDsEMG-torque coherence during these tasks differ between groups?

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CHAPTER 2

"Fatigue-induced alterations in the spatial distribution of lumbar erector spinae activity in older versus young adults"

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2.1 Abstract

Purpose: This study examined differences in the spatial distribution of lumbar erector spinae (LES) muscle activity during a submaximal isometric trunk extension contraction between older and younger adults.

Methods: Thirteen older adults (OLDER) and thirteen young adults (YOUNG) participated. High-density surface electromyography signals were recorded from the LES muscle during an isometric trunk extension task at 30% of maximal voluntary isometric force until failure. The spatial distribution of muscle activity was assessed via the x and y coordinates of the centroid of the root mean square map. Muscle fibre conduction velocity (MFCV) was calculated as a physiological index of local muscle fatigue. Force steadiness was quantified using the coefficient of variation (CoV) of force.

Results: MFCV values significantly decreased during the fatiguing task ($p < 0.001$), with the two groups showing a similar rate of decline. Significant "Time*Group" interactions were found for the centroid displacement along both the y-axis ($p = 0.017$) and the x-axis ($p = 0.006$), with OLDER showing a cranial shift of muscle activity and YOUNG a lateral shift. Endurance time was similar between groups ($p = 0.749$). The CoV was consistently higher in OLDER throughout the task ($p = 0.001$).

Conclusion: This study highlights age-related differences in LES activity adaptations to muscle fatigue, with older adults showing potentially protective, but less efficient recruitment strategies. In addition, older participants demonstrated poorer trunk force control during the task.

Keywords: Muscle fatigue · Ageing · Erector spinae · High-density EMG · Low back

2.2 Introduction

Muscle fatigue, defined as “any exercise-induced reduction in the ability of a muscle to generate force or power”, is a complex phenomenon that results from alterations in both central (i.e., excitability of the motor cortex and spinal α-motoneurons, descending corticospinal pathways) and peripheral (i.e., sarcolemmal excitability, cross-bridge cycle, muscle metabolism) factors (Gandevia 2001). Since the ability to execute a task repeatedly or for a prolonged period of time is fundamental for the successful performance of activities of daily living (Christie et al. 2011), the investigation of age-related differences in the development of fatigue has been a topic of great interest. Interestingly, despite the typical decline in muscle mass and strength with ageing, a number of studies showed increased fatigue resistance in older people as compared to young adults during isometric contractions of the lower limbs (Allman and Rice 2002; Bazzucchi et al. 2005; Kent-Braun 2009; Helbostad et al. 2010). This phenomenon has been defined as the “fatigue-paradox”, which has been attributed to progressive slower contractile properties during ageing (Hunter et al. 2016). Although most studies on ageing and muscle fatigue have focused on limb muscles, trunk muscles have received less attention. Yet, trunk muscle integrity is fundamental for functional performance during ageing (Granacher et al. 2013) and, more specifically, endurance of the trunk extensor muscles is associated with both mobility and balance performance in older adults (Kienbacher et al. 2014). However, the effects of ageing on fatigue resistance of the trunk extensors, quantified as endurance time (s), remain unclear due to conflicting results. Some authors observed greater muscle endurance in older people during a trunk extension exercise (Yassierli et al. 2007; Parreira et al. 2014), while others reported no significant differences in fatigue resistance between young and older individuals (Champagne et al. 2009; Singh et al. 2011; Tsuboi et al. 2013) or a decline in endurance with ageing (Parreira et al. 2013). Nevertheless, assessing endurance time alone is insufficient when evaluating the complex phenomenon of muscle fatigue. Including neurophysiological measurements, such as surface electromyography (sEMG), can help in interpreting muscle fatigue and how it changes with ageing. Several sEMG parameters have been extracted either in the time domain or in the frequency domain to assess myoelectric manifestations of fatigue in both upper and lower limbs during ageing (Macaluso et al. 2000; Bazzucchi et al. 2005; Duffy et al. 2012; Boccia et al. 2015). However, previous studies on trunk muscle fatigue in older participants did not always include EMG assessments (Champagne et al. 2009; Parreira et al. 2013, 2014), or, when they did, they relied on traditional bipolar sEMG (Singh et al. 2011; Tsuboi et al. 2013; Habenicht et al. 2020), which gives limited information and is limited to a small muscle region. Due to a high spatial sampling resolution, high-density sEMG (HD-sEMG) provides more thorough insight on muscle responses to fatigue. In particular, this technique provides information on the spatial distribution of motor unit activity, by detecting changes in the intensity of activity within a muscle or muscle group (Cè et al. 2020). Variation in the spatial distribution of muscle activity has been demonstrated for different muscles under various conditions, including during fatiguing contractions and contractions of increasing load, suggesting heterogeneity in the distribution of motor units or changes in the motor control strategy (i.e., inhomogeneous motor unit recruitment or substitution) (Farina et al. 2008). Interestingly, Tucker et al. (2009) reported that muscle activity of the lumbar erector spinae (LES) adapted in a nonuniform manner only during sustained fatiguing contractions and not with varying load. From a functional point of view, this adaptation is important for maintaining force output when muscle fatigue occurs (Falla et al. 2014). However, this physiological phenomenon in the LES has been investigated only in young and adult individuals with and without chronic low back pain, showing that people suffering from chronic low back pain present an altered distribution of LES activity during both isometric (Abboud et al. 2014; Sanderson et al. 2019b) and dynamic contractions (Falla et al. 2014; Sanderson et al. 2019a, 2024; Arvanitidis et al. 2021). To the best of authors' knowledge, no previous studies investigated the spatial adaptation of LES activity during fatiguing contractions in older adults. Since LES muscle is continuously exposed to repeated or prolonged activation due to its critical role in maintaining an upright posture (Falla et al. 2014) and it undergoes degenerative changes with normal ageing (Dallaway et al. 2020), the current study could provide new insights on how older age impacts the control of the LES during fatigue. Therefore, the aim of the study is to compare the spatial distribution of LES activity during a submaximal isometric trunk extension contraction between older and younger adults. Given the large number of age-related alterations in the neuromuscular system (i.e., loss of muscle fibres, motor unit remodelling, impaired contractile properties), it was hypothesised that a more homogeneous LES activity would be observed in older people compared to younger participants during the task. Based on the study by Farina et al. (2008), which found a positive correlation between the amount of change in spatial distribution of muscle activity and endurance time during an isometric contraction, a more homogeneous activation pattern is expected to result in lower endurance time.

2.3 Methods

2.3.1 Participants

Thirteen older volunteers (8 males and 5 females; mean age: 69 ± 4) and thirteen younger volunteers (8 males and 5 females; mean age: 26 ± 2) were enrolled in the study. The sample size was determined a priori based on a statistical power analysis (G*Power statistical software v.3.1.9.4) for a mixed-model ANOVA (within-between factors) (α = 0.05, power (1-β err prob) = 0.80, effect size = 0.30), as indicated by Cohen (1992) and based on the previous studies assessing the spatial distribution of muscle activity that reported a small to moderate effect size (Arvanitidis et al. 2022; Sampieri et al. 2024). Participants were recruited via information leaflets posted throughout the University of Rome "Foro Italico" and in nursing homes. Young participants (age range: 18–35 years) were included if they did not have a history of neurological or orthopaedic disorders, while older participants (age range: 65–75 years) were included if they met the criteria to be defined as “medically stable” for exercise studies (Greig et al. 1994). In addition, both young and older volunteers were excluded from the study if they had a history of chronic low back pain, current low back pain, lumbar radiculopathy, spinal surgery or spinal deformities (i.e., scoliosis, spondylolisthesis, spondylolysis), as reported by Singh et al. (2011). All participants performed recreational physical activities, but they were not involved in any kind of systematic training at the time of the study or during the six months prior (Shigaki et al. 2018). The study obtained approval from the University of Rome “Foro Italico” ethics review board (CAR 160/2023) and, after a thorough explanation of the experimental procedures, each participant provided informed written consent. The STROBE guidelines were followed for reporting this study.

2.3.2 Experimental procedures

Participants were asked to attend one experimental session lasting approximately 1 h, during which they performed a fatiguing task of the trunk extensor muscles, consisting of a modified version of the Sorensen endurance test (Champagne et al. 2009). As shown in **Fig. 1**, they were positioned in prone on a 45° Roman chair with their anterior superior iliac crests on the front edge of the pelvic pad. They had a belt placed at the level of their scapulae that was connected to a load cell (OT Bioelettronica, Turin, Italy) via a cable that was anchored to the base of the Roman chair perpendicularly to the axis of the trunk. Both the height of the pelvic pad and the length of the cable were adjusted by the investigator based on the participants’ height. Participants were first familiarised with the trunk extension movement through a warm-up consisting of 3 sets of 8 low-to-moderate intensity contractions, with 1 min of rest between sets. They were then instructed to perform a maximal voluntary isometric contraction (MVIC) test: they had to extend their trunk in 2 s until reaching a horizontal position relative to the ground, exert their maximal voluntary isometric force (MVIF) for 3 s and then return to the initial position in 2 s. Three attempts were performed with 3 min rest intervals between them. The repetition with the highest exerted force was selected to determine the target isometric force of 30% of MVIF, which represented the constant force that participants reached during the fatiguing task, as indicated by Abboud et al. (2014). During the fatiguing task, participants had their trunk unsupported in a horizontal position relative to the ground with their hands crossed over their chest (**Fig. 1**). They maintained this position for as long as possible (until voluntary exhaustion). They received constant visual feedback of the target force of 30% MVIF on a computer screen. Two red lines representing 25 and 35% of MVIF were used to define the margin of error allowed during the task. The test ended either when participants reached voluntary exhaustion or when they exceeded the lower error margin more than three times. Participants were verbally encouraged by the investigators while per forming the task. Immediately after the fatiguing task, the MVIC test was performed again to confirm the presence of muscle fatigue.

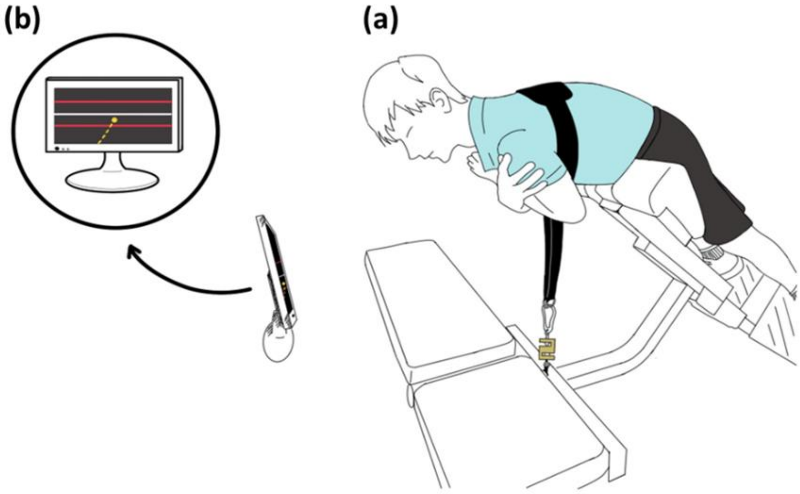


Fig. 1 Position of participants during the modified Sorensen endurance test on the Roman Chair **(a)**. Representation of the computer screen on which the target force with the margins of error were shown **(b)**.

2.3.3 HD-sEMG signal recording

The CEDE checklist was used to ensure accurate reporting of the acquisition and processing of EMG data (Besomi et al. 2024). Surface EMG signals were recorded during both the MVIC test and the fatiguing task from the right LES muscle for participants using an adhesive grid of 64 equally spaced electrodes (HD08MM1305, gold-coated, 1 mm diameter, 8 mm interelectrode distance (IED); OT Bioelettronica, Turin, Italy). The grid consisted of 13 rows and 5 columns of electrodes with a missing electrode in the upper right corner. After identifying the perimeter of the LES by manual palpation, the participant's skin was shaved and gently abraded using abrasive paste (Skin Prep Gel NUPREP) to ensure the best conductivity of HD sEMG signals. As reported by Falla et al. (2014), the grid was placed ~2 cm lateral to the lumbar spinous process mid-point, covering the low back approximately from L5 to L2 (**Fig. 2**). The grid was secured over the skin with a double-sided adhesive foam layer with holes adapted to the HD-sEMG grid (SpesMedica, Battipaglia, Italy). The holes of the foam layer were filled with conductive paste (SpesMedica, Battipaglia, Italy) to provide skin–electrode contact. The reference electrode was placed on the ulnar styloid process of the left wrist with a moistened strap. The HD-sEMG signals were amplified using a multichannel amplifier (Sessantaquattro, OT Bioelettronica, Turin, Italy), sampled at 2000 Hz, and converted to digital data.

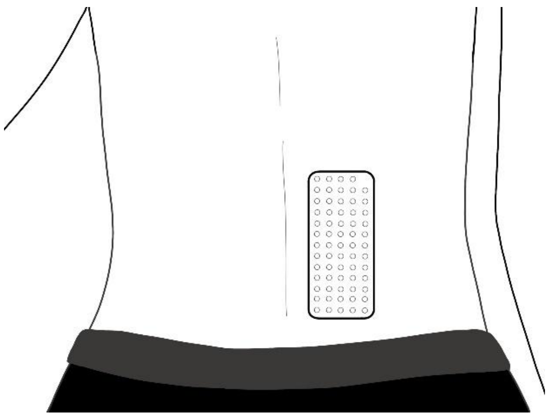


Fig. 2 Grid of electrodes positioned over the right lumbar erector spinae (LES) muscle of participants.

2.3.4 HD-sEMG signal processing

The HD-sEMG signals were analysed offline using a custom-made Matlab script (MATLAB 2022b, Mathworks Inc., Natick, MA, USA). Fifty-nine adjacent bipolar channels were obtained from the grid (twelve longitudinal bipolar recordings in each column, except for the far right with eleven electrode pairs) (Arvanitidis et al. 2021). HD-sEMG signals were bandpass filtered (1st order, 3 dB bandwidth, 10–500 Hz). Root mean square (RMS) values were computed from each bipolar channel. The signal quality of each channel was visually inspected, and channels with low signal-to-noise ratio were removed. Because the removal rate was less than 15% in the present study, the EMG variables could be reliably estimated (Arvanitidis et al. 2021). The RMS values averaged over the 59 bipolar channels were measured (RMSmean). The RMSmean values were normalised to the maximum RMS (RMSmax) extracted from the MVIC test performed before the fatiguing task to allow statistical comparisons between participants. To characterise the spatial distribution of LES muscle activity during the fatiguing task, a topographical map of the RMS values was generated. The x- and y-coordinates of the centroid of the topographical map were estimated for the medial-lateral and cranial-caudal direction, respectively. In addition, the modified entropy of the RMS values was calculated according to Farina et al. (2008) as follows (**Eq. 1**):

$$\text{Entropy} = \sum_{i=1}^{59} p^2(i) \log_2 p^2\,(i)$$

Equation 1 Entropy is computed from the normalised squared RMS values, with $p^2(i)$ defined as the squared RMS value at electrode i normalised by the sum of the squared RMS values across the 59 electrodes.

The entropy of a set of M values is maximal when all values are identical (i.e., a uniform distribution), taking a value of $\log_2 M$, and minimal when it equals zero. In this context, entropy represents the degree of homogeneity of muscle activity, with higher values indicating a more homogeneous distribution of activity over the grid (Farina et al. 2008). Lastly, muscle fibre conduction velocity (MFCV), which is a well-known index of local muscle fatigue (Merletti et al. 1990), was estimated using an in-built function of OTBioLab+ (version 1.6.0), the same software used for data recording. HD-sEMG signals were visually inspected and the three consecutive channels with the highest signal quality were chosen. The criteria for channel selection were the clearest propagation of motor unit action potential (MUAP) along the cranial-to-caudal direction and MFCV values within the physiological range (2–6 m/s) (Nuccio et al. 2020; Brouwer et al. 2022). The correlation coefficient was calculated between the selected channels (Brouwer et al. 2022). All EMG variables were calculated in five equal and consecutive time epochs during the fatiguing task, corresponding to the 0, 25, 50, 75, and 100% of the endurance time of each participant.

2.3.5 Force analysis

During offline analysis, the force signal was converted to torque (Nm) by multiplying the MVIF value (kg) by 9.8 and by the lever arm, defined as the distance from the hip joint axis of rotation to the point where the belt was positioned at the level of the scapulae. Low-pass filtered with a cutoff frequency of 15 Hz (4th order, zero-lag, Butterworth) was applied. During the fatiguing task, the amplitude of force fluctuations was assessed in relative terms as the coefficient of Variation (CoV), reported as percentage (**Eq. 2**). It was calculated in the same time epochs (0, 25, 50, 75, and 100% of the endurance time) used for the HD-sEMG analysis. The custom-made Matlab script was designed to plot the force exerted by each participant and, after identifying the steady part of the contraction, the starting and ending points required for the analysis were chosen.

$$CoV = \frac{force\ standard\ deviation\ (SD)}{force\ mean} \times 100$$

Equation 2 Coefficient of Variation (CoV) corresponding to the amplitude of force fluctuations and computed as the relative ratio between the standard deviation (SD) and mean value of the force output.

2.3.6 Statistical analysis

Statistical analysis was performed using IBM SPSS 24.0 (IBM Corp., Armonk, NY, United States). The Shapiro–Wilk test was used to check normality of the data. When the normality was not assumed, nonparametric tests were applied. Independent t-tests were performed to compare young and older participants for the following variables: anthropometric measures (body mass, height, and BMI), MVIF values both before and after the fatiguing task (MVIF-pre and MVIF-post, respectively), MVIF decline, endurance time, and mean exerted force during the task. MVIF decline was defined as the difference between the MVIF-post and MVIF-pre values divided by the corresponding MVIF-pre values and expressed in percentage. The mean exerted force was also expressed as a percentage of the MVIF. A mixed-model ANOVA was used to detect potential differences in EMG variables (RMSmean, MFCV, x and y coordinates of the centroid) and CoV between the two groups and throughout the fatiguing task. The Aligned Rank Transform (ART) ANOVA (nonparametric approach to factorial ANOVA) was applied to analyse entropy due to its not normal distribution. “Time” represented the within-participant factor with five levels (0, 25, 50, 75, and 100% of the endurance time), and “Group” represented the between-participant factor (‘YOUNG’ referred to young participants, while “OLDER” referred to older participants). When a significant main effect or interaction was found, paired t-tests were used for post-hoc analysis. For the ART ANOVA, pairwise t-tests were performed on the aligned and rank-transformed data rather than on the raw values. For all statistical tests, significance level (α) was set to 0.05, with a Haulm-Bonferroni correction for multiple post hoc comparisons. Mauchly test was used to check sphericity of the data for the parametric ANOVA analysis. The data are reported as mean ± Standard Deviation (SD) for parametric data and as median and interquartile range (IQR) for nonparametric data.

2.4 Results

2.4.1 Anthropometric characteristics and endurance time

Independent t-tests showed no significant differences in body mass, height and BMI between YOUNG and OLDER groups ($p > 0.05$). In addition, similar levels of endurance time were observed in the two groups ($p > 0.05$). The results for these variables are shown in **Table 1**.

Table 1 Participants' anthropometric characteristics and endurance time (mean ± SD).

Characteristic	YOUNG (n=13)	OLDER (n=13)	*p value*
Body mass (kg)	70.62 ± 15.19	70.23 ± 8.17	0.937
Height (m)	1.73 ± 0.13	1.72 ± 0.08	0.895
BMI (kg/m^2)	23.32 ± 2.44	23.58 ± 1.80	0.762
Endurance time (s)	104.32 ± 38.17	99.74 ± 34.03	0.749

2.4.2 HD-sEMG activity

The mixed ANOVA revealed a significant main effect of "Time" for the RMSmean values ($F = 10.186$, $\eta p^2 = 0.298$, $p < 0.001$), while no significant main effect of "Group" ($F = 2.423$, $\eta p^2 = 0.092$, $p = 0.133$) or "Time*Group" interaction ($F = 0.845$, $\eta p^2 = 0.034$, $p = 0.439$) were observed. Post hoc analysis showed a significant increase from 0 to 25% ($p < 0.001$), 0 to 50% ($p < 0.001$), 0 to 75% ($p = 0.003$), 25 to 50% ($p = 0.01$), and a significant decrease from 50 to 100% ($p < 0.001$) and 75 to 100% ($p = 0.001$), as illustrated in **Fig. 3a**. MFCV values showed a significant main effect of "Time", decreasing in both groups over time ($F = 75.636$, $\eta p^2 = 0.759$, $p < 0.001$), with each time epoch differing from the others ($p < 0.001$) (**Fig. 3b**). Neither a significant main effect of "Group" ($F = 1.764$, $\eta p^2 = 0.068$, $p = 0.197$) nor a significant "Time*Group" interaction ($F = 0.177$, $\eta p^2 = 0.007$, $p = 0.752$) was observed. The average correlation coefficient between selected channels for calculating MFCV was 0.67 ± 0.17 for YOUNG and 0.54 ± 0.18 for OLDER.

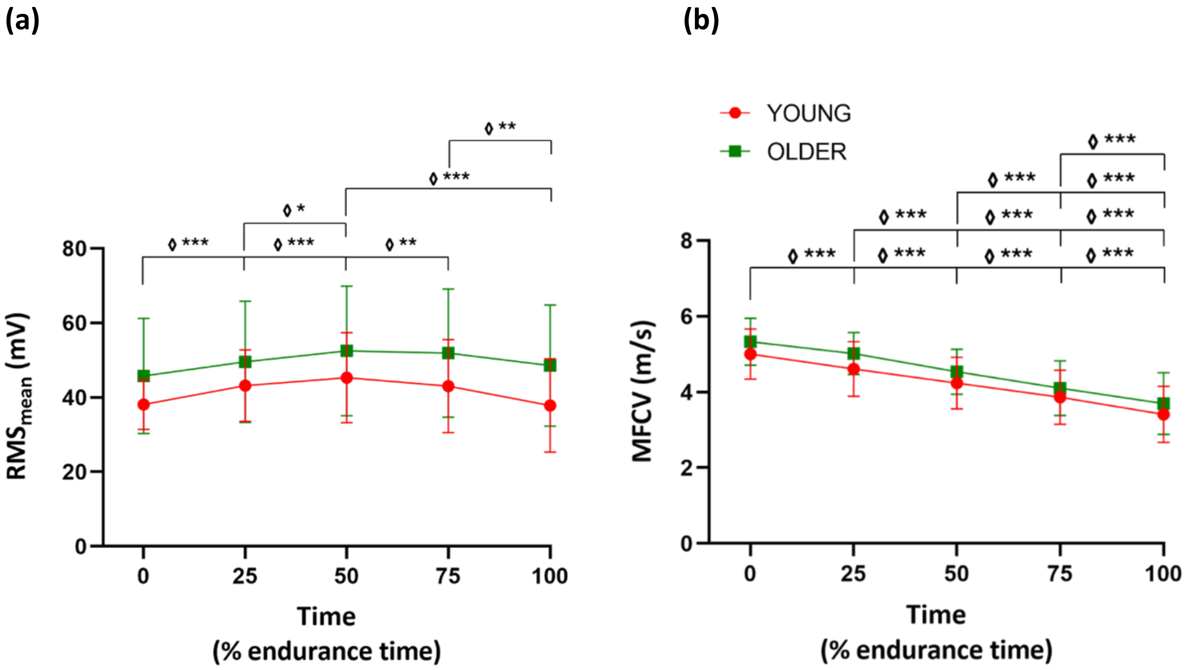


Fig. 3 Root mean square (RMSmean) (a) and muscle fibre conduction velocity (MFCV) (b) values averaged over the grid for 0, 25, 50, 75, and 100% of endurance time of both groups. Data are reported as mean ± SD. ◊ significant main effect of "Time"; *$p < 0.05$; **$p < 0.01$; ***$p < 0.001$.

For the topographical map of the RMSmean values, a significant main effect of "Time" ($F = 3.476$, $\eta p^2 = 0.127$, $p = 0.044$) and a significant "Time*Group" interaction ($F = 6.043$, $\eta p^2 = 0.201$, $p = 0.006$) were found for centroid displacement along the x axis. However, no significant main effect of "Group" was observed ($F = 0.007$, $\eta p^2 = 0.000$, $p = 0.934$) (**Fig. 4a**). Post hoc analysis of the main effect of "Time" did not reveal any significant differences. In contrast, post hoc analysis of the "Time*Group" interaction indicated no significant displacement of the centroid over time along the x axis in the OLDER group ($p > 0.05$), while the YOUNG group exhibited a lateral dis placement of the centroid, especially between 0 and 100% time epochs ($p = 0.004$). A significant "Time*Group" interaction ($F = 5.111$, $\eta p^2 = 0.176$, $p = 0.017$) was also shown for the displacement of the centroid along the y axis, while neither a significant main effect of "Time" ($F = 0.428$, $\eta p^2 = 0.018$, $p = 0.603$) nor a significant main effect of "Group" ($F = 3.081$, $\eta p^2 = 0.114$, $p = 0.092$) were observed (**Fig. 4b**). Post hoc analysis revealed a cranial shift of the centroid in OLDER from 0 to

25% (p = 0.003), 0 to 75% (p = 0.003), 0 to 100% (p = 0.004), 50 to 75% (p = 0.002), and from 50 to 100% (p = 0.006), whereas no significant displacement was found in YOUNG (p > 0.05).

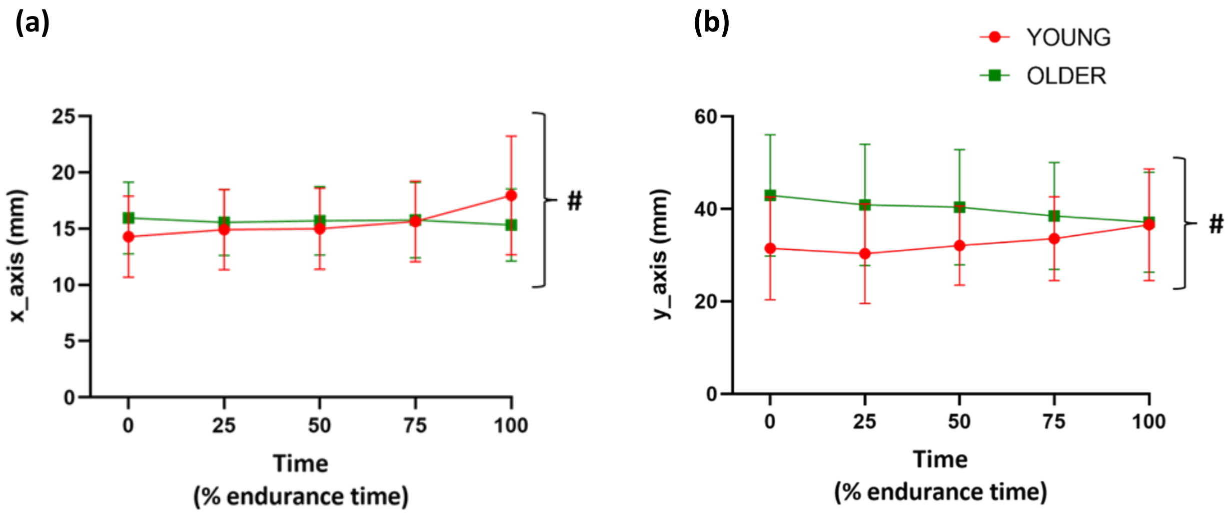
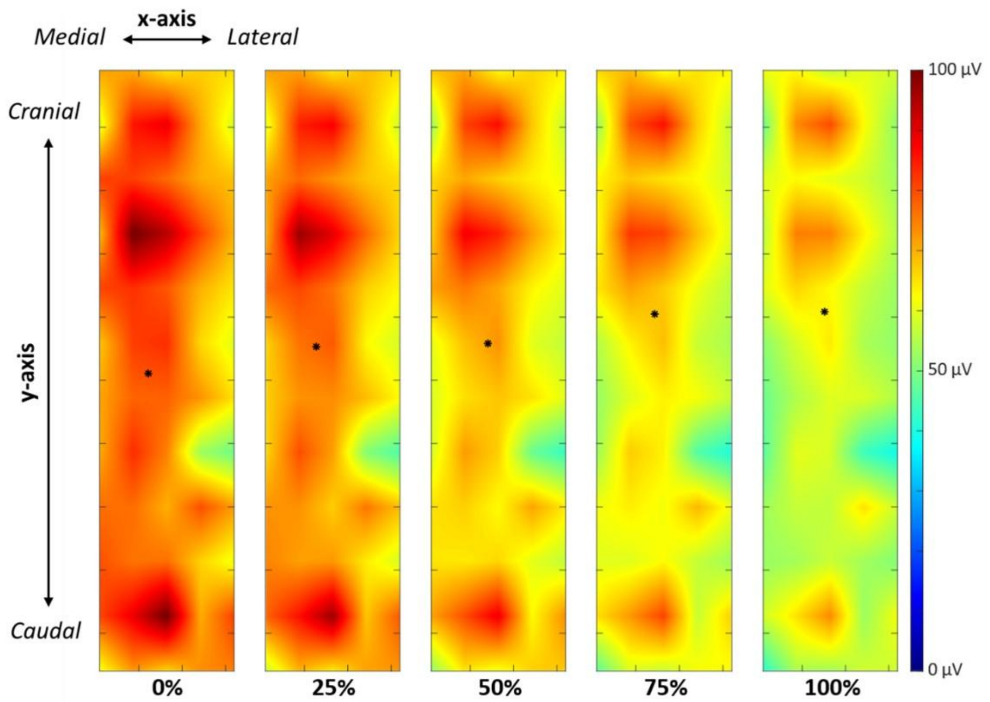


Fig. 4 Displacement of the centroid of lumbar erector spinae (LES) muscle activity along the x axis (a) and y axis (b) for 0, 25, 50, 75, and 100% of endurance time of both groups. Data are reported as mean ± SD. # significant "Time*Group" interaction (p < 0.05).

Figure 5 shows the topographical maps for two representative participants (one older (**Fig. 5a**) and one young (**Fig. 5b**) participant) across the time epochs considered.

(a)



(b)

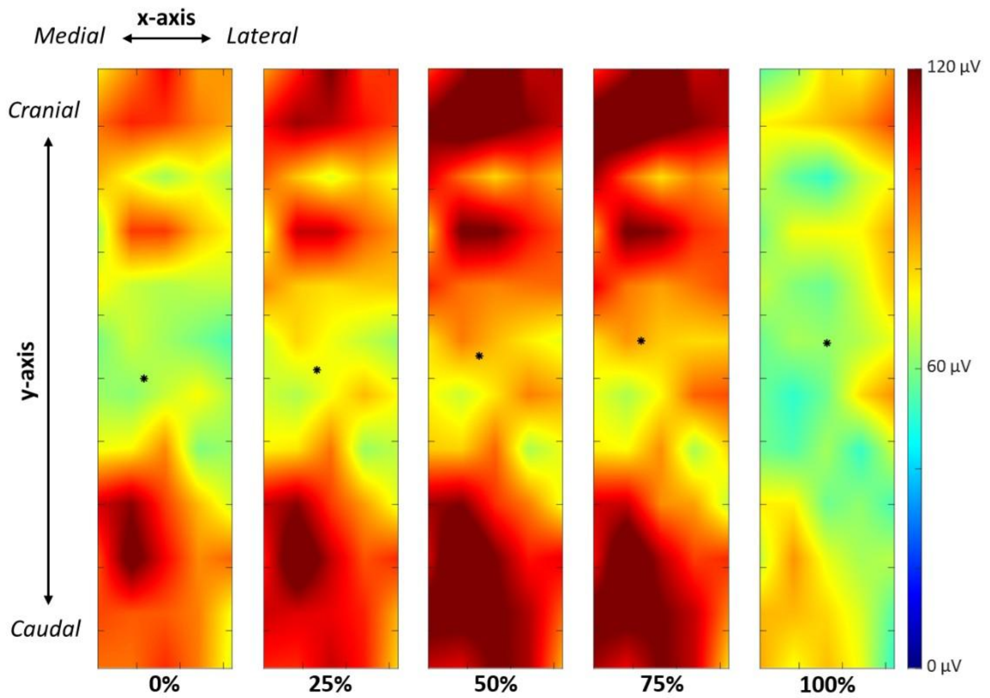


Fig. 5 Representative topographical maps of lumbar erector spinae (LES) muscle activity for one older (a) and one young (b) participant at 0, 25, 50, 75, and 100% endurance time. The black dot in the maps indicates the centroid of

muscle activity. The amplitude of each high-density surface EMG (HD-sEMG) channel is color-coded according to the bar on the right, with red colour indicating regions of the LES muscle with higher muscle activity.

Lastly, no significant main effects of "Time" ($F = 0.060$, $\eta p^2 = 0.002$, $p = 0.953$), "Group" ($F = 0.118$, $\eta p^2 = 0.005$, $p = 0.734$) and no significant "Time*Group" interaction ($F = 0.031$, $\eta p^2 = 0.001$, $p = 0.978$) were observed for the entropy values of the topographical map (**Fig. 6**).

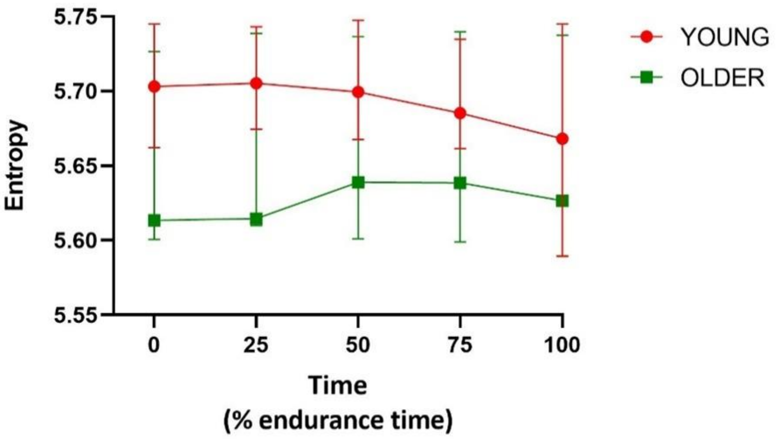


Fig. 6 Entropy of the root mean square (RMS) values for 0, 25, 50, 75, and 100% of endurance time of both groups. Data are reported as median (IQR).

2.4.3 Torque variables

The t-test analysis performed on the MVIF-pre and MVIF-post values showed significant differences between OLDER and YOUNG ($p < 0.05$), as illustrated in **Table 2**. However, MVIF decline was not different between the two groups ($p > 0.05$). No differences were found in the mean exerted force during the task between OLDER and YOUNG ($p > 0.05$).

Table 2 Maximal Voluntary Isometric Force (MVIF) pre, MVIF post, MVIF decline, and mean force values of both groups reported as mean ± SD.

Characteristic	YOUNG (n=13)	OLDER (n=13)	*p value*
MVIF pre (Nm)	262.02 ± 100.41	166.53 ± 73.16	0.011
MVIF post (Nm)	210.70 ± 90.97	127.19 ± 61.80	0.011
MVIF decline (%)	-20.45 ± 8.25	-22.70 ± 17.67	0.681
Mean force (%)	29.42 ± 0.27	29.04 ± 0.65	0.070

The CoV showed a significant main effect of "Time" ($F = 39.418$, $\eta p^2 = 0.622$, $p < 0.001$), increasing significantly in both groups from 0 to 75% ($p = 0.001$), 0 to 100% ($p < 0.001$), 25 to 75% ($p < 0.001$), 25 to 100% ($p < 0.001$), 50 to 75% ($p = 0.001$), 50 to 100% ($p < 0.001$), and 75 to 100% ($p < 0.001$) (**Fig. 7**). Additionally, a significant main effect of "Group" was observed ($F = 13.628$, $\eta p^2 = 0.362$, $p = 0.001$), with the OLDER group having a consistently higher CoV for all time epochs. Instead, the "Time*Group" interaction was not significant ($F = 1.366$, $\eta p^2 = 0.054$, $p = 0.264$).

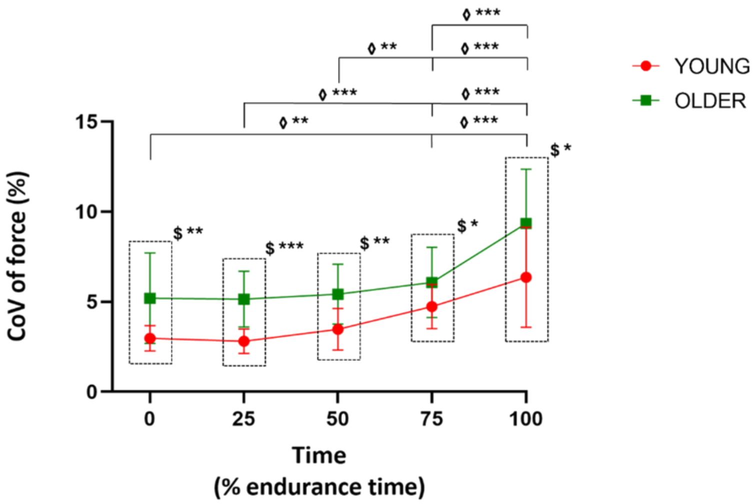


Fig. 7 Coefficient of Variation (CoV) of force for 0, 25, 50, 75, and 100% of endurance time of both groups. Data are reported as mean ± SD. ◊ significant main effect of "Time"; $ significant main effect of "Group"; *p < 0.05; **p < 0.01; ***p < 0.001.

2.5 Discussion

This study compared the spatial distribution of LES muscle activity between younger and older adults as they per formed submaximal isometric trunk extension until voluntary exhaustion. Contrary to our hypothesis, both older and young participants showed a spatial reorganisation of LES activity with fatigue. Interestingly, the centroid of muscle activity shifted cranially in older participants, whereas it shifted laterally in young controls.

In accordance with the previous studies assessing trunk extensor strength (Singh et al. 2011), MVIF values of older participants were significantly lower compared to their younger counterparts.

However, the two groups showed a similar decline in MVIF values after the fatiguing task, suggesting that they reached comparable levels of fatigue. From a physiological perspective, the presence of muscle fatigue is confirmed by the marked decrease in MFCV values that both older and young participants showed during the fatiguing task. MFCV is an index of local muscle fatigue (Merletti et al. 1990; Farina et al. 2007) that is considered to be more reliable than spectral variables to assess the rate of fatigue (Boccia et al. 2015). MFCV typically decreases during sustained muscle contractions due to alterations at a peripheral level such as accumulation of metabolites and changes in muscle fibre membrane excitability and propagation (Milner-Brown and Miller 1986; Stewart et al. 2011). In addition, as muscle fatigue progressed during the task, the decrease in MFCV was accompanied by an increase in RMS amplitude. This probably results from an intensified excitatory neural drive to the LES, which reflects the recruitment of additional motor units to compensate for the fatigue-induced decline in muscle contractile efficiency and maintain the required target force (Carpentier et al. 2001). However, RMS amplitude returned to the initial values at the end of the task (100% of endurance time). Slowing of muscle contractile properties during submaximal prolonged contractions has been previously associated with a reduction in motor unit discharge rate, which could partially account for the observed decrease in RMS amplitude close to failure (Garland and Gossen 2002). Nevertheless, given the nature of the task in the present study, the recruitment of synergistic muscles (i.e., hip extensors) as fatigue progressed in the LES (Rossato et al. 2022) likely played a primary role in the return of RMS amplitude to baseline levels. Although participants' position on the Roman Chair (i.e., inclination of the trunk) was chosen to maximise isolation of the trunk extensor muscles (Demoulin et al. 2006; Champagne et al. 2009), activation of synergistic muscles remains plausible.

Both groups showed a displacement of the centroid of the topographical EMG amplitude map during the fatiguing task, indicating a redistribution of LES activity with muscle fatigue. Although the EMG centroid displacement is of a few millimetres, it reflects relatively large changes in EMG amplitude distribution, indicating substantial alterations in muscle activation along and/or across muscle fibres, as previously suggested (Falla and Gallina 2020). This redistribution of activity may represent a mechanism to maintain the target force and prolong the contraction despite altered afferent feedback (i.e., fatigue), preventing overload of specific muscle regions (Farina et al. 2008). This mechanism is likely caused by changes in motor control strategies such as space-dependent motor unit recruitment, modulation of discharge rate or motor unit substitution (Farina et al. 2008). Interestingly, LES activity was redistributed differently between the two groups, with older participants showing a cranial shift in muscle activity, unlike younger participants. Abboud et al. (2020) have recently demonstrated the presence of cranial and caudal motor unit territories within the longissimus pars lumborum muscle, which can be selectively activated by the central nervous system during voluntary movements. The more cranial muscle activation patterns observed in older adults may represent a protective mechanism aimed at redistributing the load toward the upper lumbar regions. This could help reduce compressive and shear forces acting on the intervertebral discs of the lower lumbar spine, particularly at L4–L5 level, which commonly show greater degeneration with ageing (Hicks et al. 2009). However, the previous studies reported that a cranial shift of LES activity may indicate less favourable contraction from a biomechanical perspective, as muscle fibres in the cranial regions primarily extend the upper lumbar vertebrae rather than the entire lumbar spine (Sanderson et al. 2019b, a). Overall, although the strategy adopted by older adults may represent a protective or compensatory mechanism, it may be inefficient, as it could limit the ability of the LES to generate force optimally. In contrast, the motor control strategy observed in younger participants, characterised by comparable activation of both caudal and cranial regions of the LES, may be more efficient, as it distributes the load across a larger number of muscle fibres. This may help minimise the effort required to sustain the contraction (Sanderson et al. 2019b; Arvanitidis et al. 2021). In addition, the younger group showed a lateral shift of the centroid with fatigue, suggesting a redistribution of mechanical loads within the LES. Abboud et al. (2023) observed increased activation in the lateral regions of the LES when postural control was challenged, particularly in response to external trunk perturbations. This activation strategy may be associated with a greater mechanical advantage in cases of instability. Therefore, the more lateral activation observed in young participants may have been a response to counteract reduced trunk stability due to fatigue.

It is well-established that muscle fatigue negatively affects force steadiness, resulting in increased force fluctuations (Salomoni and Graven-Nielsen 2012). Both young and older participants showed a marked increase in the CoV of force from the beginning to the end of the fatiguing task. Force control of trunk extensor muscles is an important aspect of physical function as these muscles are continuously exposed to prolonged contractions and they play a fundamental role in stabilising the spine (Granacher et al. 2013). Notably, the CoV was consistently higher in older participants compared to young controls, suggesting that they had poorer trunk extensor muscle control during the task. The lower force steadiness that we found with advanced age is in accordance with previous studies assessing other muscles (Bazzucchi et al. 2004; Enoka and Farina 2021). However, to the best of authors' knowledge, only one study has assessed force steadiness of the trunk extensor muscles of older individuals so far, revealing poorer muscle force control at 10% of peak torque with advanced age (Porto et al. 2020). Using muscle coherence analysis, Castronovo et al. (2018) demonstrated that the reduced force steadiness in older individuals is related to the synaptic input received by motoneurons. Particularly, the age-related increase in the fluctuations of the low-frequency common synaptic input negatively affects force accuracy during submaximal isometric contractions. Future studies should apply muscle coherence analysis on trunk muscles to better understand the mechanisms underlying force control in these muscles during ageing.

Despite greater impairments in force control among older participants, the two groups had comparable endurance times during the fatiguing task. This result is in accordance with some previous studies that assessed the endurance of the lumbar extensor muscles of older adults (Champagne et al. 2009; Singh et al. 2011; Tsuboi et al. 2013). A number of studies reported greater fatigue resistance in other muscles with ageing, which is usually attributed to a progressive age-related shift toward slower contractile properties, as fast twitch fibres are more affected by ageing. However, this may not apply to trunk muscles, which predominantly consist of slow-twitch fibres (approximately 60–70%) (Jørgensen et al. 1993). This is supported by the similar rate of decrease in MFCV values that young and older participants showed in this study. Indeed, based on the different membrane proper ties of slow-twitch and fast-twitch muscle fibres (Milner Brown and Miller 1986), the rate of change of MFCV has been associated to muscle fibre composition (Boccia et al. 2015). For example, Merletti et al. (2002) demonstrated lower decrements of MFCV in older adults compared to young controls during submaximal isometric contractions of the biceps brachii, suggesting higher proportion of slow-twitch muscle fibres, and thus lower muscle fatigability, during ageing. However, this pattern was not observed in our study.

Lastly, although a spatial adaptation occurred in both groups during the fatiguing task, entropy values of the topographical EMG amplitude map did not change over time. Lower entropy values are typically associated with spatial reorganisation of muscle activity across muscle fibres (Farina et al. 2008). However, similar results were reported by Arvanitidis et al. (2021), who observed no change in entropy values of the HD-sEMG map despite preferential activation of certain muscle regions over others during an isokinetic fatiguing protocol for the trunk muscles. One possible explanation is that, although the shift in the centroid of LES activity was significant, the detection sites may not have been enough to induce a decrease in entropy. For example, Sanderson et al. (2019a), who reported a spatial reorganisation of erector spinae activity along with a significant decrease in entropy values in patients with chronic low back pain, used two HD-sEMG grids of 64 electrodes covering the thoracolumbar erector spinae. In addition, the task employed in this study was isometric and less com plex compared to those used in some previous studies. The higher complexity of lifting (Sanderson et al. 2019a) or rowing (Martinez-Valdes et al. 2019) which was assessed in other studies likely engaged different trunk muscle regions to a greater extent, increasing the heterogeneity of the EMG amplitude map. The results of the present study further demonstrate that entropy only provides a general measure of the heterogeneity of HD-sEMG map, whereas assessing the centroid of muscle activity gives more detailed information about the specific distribution of activity.

In the present study, some limitations should be considered. First, given the nature of the task, it is plausible that muscles other than the LES muscle, such as hip muscles, were activated during both the MVIC tests and fatiguing protocol. To address this, future studies should consider assessing synergistic muscles to evaluate their activation levels or employing a specific dynamometer for trunk muscles to reduce the involvement of other muscle groups. However, to minimise this issue in the current study, participants were instructed to focus on activating their trunk muscles and to avoid compensatory movements. Additionally, investigators constantly monitored participants to ensure proper technique during the test. Second, due to participants' position, those with higher body mass may have supported more weight from their upper body, even if exerting the standardised relative effort (30% MVIF). However, as there were no significant differences in anthropometric characteristics between the groups (Table 1, Section 2.4.1), this factor may not have substantially affected the results. Lastly, the correlation coefficient between selected channels used to estimate MFCV was lower in older participants compared to the younger adults. Although a correlation coefficient above 0.8 has typically been used, Del Vecchio et al. (2018) demonstrated that reliable MFCV estimation is possible with a threshold of ≥0.5. Additionally, Brouwer et al. (2022) reported that in a complex muscle group like the back muscles, reliable MFCV estimates can be achieved even with a minimum correlation coefficient between 0.30 and 0.40. Therefore, we can assume that reliable MFCV values are reported in our study. However, since this is, to the best of authors' knowledge, the first study to assess MVCF in the trunk muscles of older adults, future studies are warranted to confirm our findings.

2.6 Conclusion

This is the first study to investigate the neuromuscular adaptations of the LES muscle during fatiguing contractions in older versus younger adults. Older participants showed a cranial shift of the centroid of muscle activity, unlike young controls, likely reflecting a protective but less efficient motor control strategy. In addition, the force control was worse in older participants during the fatiguing task, whereas endurance time was similar between groups. Reduced force steadiness, particularly in the presence of fatigue, may impair functional performance during daily activities. Future research should focus on exploring the neuromuscular mechanisms underlying deficits in trunk force control with older age, as well as identifying interventions to address this.

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CHAPTER 3

"The effects of ageing on fatigue and endurance of the spinal extensor muscles: a systematic review and meta-analysis"

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3.1 Abstract

The endurance capacity of the spinal extensor muscles plays a key role in maintaining spinal function. This systematic review and meta-analysis aims to synthesise current evidence on how ageing influences fatigue of the spinal extensor muscles, addressing the inconsistent findings reported across existing studies. Medline, EMBASE, PubMed, Web of Science and CINAHL Plus databases were searched from their inception to 28 June 2025. Cross sectional studies assessing fatigue of the spinal extensor muscles of healthy older adults (>60 years) versus younger adults were included. Methodological quality was evaluated using the Appraisal Tool for Cross Sectional Studies and the GRADE approach was applied to assess the certainty of evidence. Results were synthesised using both narrative and quantitative approaches. A random-effects meta-analysis was conducted for endurance time. The PRISMA guidelines were followed for reporting. Of the 1253 records screened, 13 studies were included, with 9 contributing to the meta-analysis. The meta-analysis, supported by moderate certainty of evidence, revealed a significant reduction in endurance time of the back extensor muscles during sustained isometric con tractions in older adults compared to younger controls (MD = −41.31; 95% CI: −64.04; −18.57). The remaining outcomes were synthesised narratively: electromyograph (EMG) related measures showed mixed findings, likely due to methodological variability across studies, while force decline and Borg ratings were reported in only a few studies. This systematic review revealed that the endurance time of the back extensor muscles is reduced during isometric tasks in older adults. However, inconsistent EMG findings limit our understanding of the neuromuscular mechanisms underlying this decline.

Keywords Ageing · Muscle fatigue · Endurance · Spinal extensor muscles · Back extensor muscles

3.2 Introduction

Optimal functioning of the spinal extensor muscles, which include the lumbar, thoracic and cervical (neck) extensors, is essential for spinal posture and function. As these muscles are usually exposed to sustained or repetitive low-level contractions to maintain upright posture, previous studies have emphasised the importance of their endurance capacity (Das 2016; Ghamkhar and Kahlaee 2019). Thus, muscle fatigue, defined as a reduction in the force/power-generating capacity of skeletal muscles in response to prolonged or repeated activity (Gandevia 2001), has been widely investigated in both back and neck muscles, across healthy and clinical populations. For instance, it has been demonstrated that fatigue of the back extensor muscles can impair postural control during both static (Lin et al. 2009; Johanson et al. 2011) and dynamic tasks (Granata and Gottipati 2008). Additionally, reduced back muscle endurance has been identified as a predictor of long-term back-related disability (Enthoven et al. 2003). Similarly, fatigue of neck extensor muscles has been associated with altered cervical position sense (Reddy et al. 2012), and decreased endurance time has been previously reported in individuals with chronic neck pain (Reddy et al. 2021).

Given the critical role of spinal extensor muscles for spinal health, their function has also been investigated in the context of ageing. Previous studies have demonstrated that both paravertebral and cervical extensor muscles undergo age-related degenerative changes, including muscle atrophy and increased fat infiltration (Okada et al. 2011; Dallaway et al. 2020). Interestingly, Beauchamp et al. (2016) reported that trunk extensor endurance is among the key physical factors predicting participation in life roles in older adults. Improvements in trunk extension endurance have also been associated with enhanced balance in community-dwelling older adults with mobility limitations (Suri et al. 2011). Moreover, as the prevalence of neck pain increases with age and has a substantial impact on quality of life (Safiri et al. 2020), improving neck extensor endurance may be beneficial for older populations to prevent and/or reduce pain (Cheng et al. 2015; Teichert et al. 2023).

Several studies have investigated how the endurance of spinal extensor muscles changes with ageing. Some authors have observed lower endurance time in older people when performing the Ito test (Han et al. 2024) and the Biering-Sørensen back endurance test (Adedoyin et al. 2011). However, other studies have reported either increased endurance time during dynamic back extensions in the elderly (Parreira et al. 2014) or no significant differences in trunk extension endurance between older and younger individuals during a modified Sørensen test (Champagne et al. 2009; Bašič et al. 2013). Additionally, some previous studies have employed surface electromyography (sEMG) to assess the myoelectric manifestations of fatigue in spinal muscles, typically by analysing the decline in median frequency (MF) and/or mean power frequency (MPF) over time. While some authors have reported a smaller decline in MF in older participants during sustained isometric back extensions (Kienbacher et al. 2014), others found no significant differences during a back extension exercise using the Roman Chair (da Silva et al. 2015), or reported a reduced decline only in older men, but not in women (Tsuboi et al. 2013).

In light of inconsistent findings across the literature, this systematic review and meta-analysis aims to synthetise and critically evaluate existing evidence on whether the development of fatigue in the spinal extensor muscles and endurance time differs in older versus younger adults.

3.3 Methods

This review is reported in line with the 2020 guidelines of Preferred Reporting Items for Systematic reviews and Meta-Analyses (PRISMA) (**Appendix - Supplemental file 1**) (Page et al. 2021). The protocol for this review was registered on the International Prospective Register of Systematic Reviews (PROSPERO; CRD42025645295) on 31 January 2025.

3.3.1 Eligibility criteria

The PICOS framework, which includes Population, Intervention, Comparison, Outcomes and Study design, was used to define the inclusion criteria for this systematic review. However, the original term "Intervention" was replaced with "Indicator" due to the nature of potentially eligible studies, as reported by previous similar systematic reviews (Devecchi et al. 2021; Arvanitidis et al. 2024).

3.3.1a Population

Studies were considered eligible for inclusion if they involved physically independent older adults aged 60 years and above. Although the age cutoff for defining an "older" person may vary across countries, the 60+ threshold was selected based on the United Nations' definition (UNHCR 2025). Additionally, studies were required to include a control group of young and/or middle-aged individuals as comparators. Only studies involving healthy participants were considered, defined as individuals free from musculoskeletal dis orders, neuro-systemic degenerative diseases, chronic cardiovascular or respiratory system diseases. Participants also had to be free from a history of chronic spinal pain, current spinal pain, spinal surgery, or spinal deformities. Lastly, there were no restrictions in terms of gender and/or ethnicity.

3.3.1b Indicator

Eligible studies were those assessing muscle fatigue in the spinal extensor muscles and/or endurance, including lumbar, thoracic, and neck extensor muscles. Studies involving voluntary fatiguing contractions performed either for a fixed duration or until task failure were included. The tasks involved the use of a specific dynamometer for trunk muscles or a standardised test for spinal extensor endurance, such as the Biering-Sørensen test or the Ito test. The Biering–Sørensen test and its variations (i.e., modified Sørensen test on a Roman Chair) and the Ito test are standardised isometric tests designed to assess the endurance of the back extensor muscles

(Demoulin et al. 2006). In these tests, participants are positioned prone and instructed to lift the upper body to maintain a horizontal posture for as long as possible or for a predetermined duration. These tests differ primarily in the type of support (e.g., Roman chair or bench), fixation methods (e.g., lower body stabilisation) and arm positioning. All types of contractions were included (i.e., isometric or dynamic), measured both at an absolute force/ torque level or relative to the individual's maximal voluntary contraction (MVC).

3.3.1c Comparison

Eligible studies included comparisons of specific outcome variables (e.g., endurance time, EMG parameters) between groups, such as older versus younger participants or older versus middle-aged participants.

3.3.1d Outcomes

Outcome variables included any measure related to muscle fatigue development and endurance. For this purpose, both mechanical variables, such as endurance time and decline in force values between pre- and post-fatigue assessments, and neuromuscular variables, such as MF or MPF of the EMG signals and muscle fibre conduction velocity (MFCV), were considered.

3.3.1e Study design

Only cross-sectional studies were included in this systematic review, as this design is most commonly used to address the research question. Therefore, non-original literature, such as systematic and narrative reviews as well as other research designs such as randomised controlled trials, were excluded. To minimise the risk of bias, studies in all languages were included in the search. However, non-English studies were ultimately excluded due to time and resource constraints.

3.3.2 Information sources

Electronic searches were performed on Medline (Ovid Interface), EMBASE (Ovid Interface), Pub Med, Web of Science (Clarivate Analytics) and CINAHL Plus (EBSCO Interface) databases from their inception to 28 June 2025. Hand searching of key journals—including Journal of Physiology, Journal of Neurophysiology, Journal of Electromyography and Kinesiology, European Journal of Applied Physiology, Journal of Applied Physiology, Muscle & Nerve—was also conducted. Additionally, the reference lists of all included papers were checked manually to identify any additional relevant studies that could have been missed during the search.

3.3.3 Search strategy

The search was conducted by the lead author (MP) without any restrictions in terms of date, geographical area and language. Search strategies were tailored for each database, and Medical Subject Headings (MeSH) were used when appropriate to optimise the search process. This approach combined MeSH terms and free-text keywords. Although the search was adapted for different databases (e.g., syntax), consistency was ensured. The full electronic search strategies for all databases are reported in **Appendix - Supplemental file 2**.

3.3.4 Selection process

All search results were imported in EndNote 20 (Clarivate Analytics) by the lead author (MP). Duplicates were identified and automatically removed by the software. All remaining references were then imported into Covidence (Veritas Health Innovation, Melbourne, Australia), where the title/abstract and full-text screening processes were conducted. Titles and abstracts of the studies were independently screened by two reviewers (MP and MA) using a pretested screening form. Studies were categorised as eligible, ineligible or doubtful. Doubtful studies were discussed between the two reviewers, and a third reviewer (DF) was involved in resolving any disagreements or uncertainties. After the initial screening phase, eligible studies underwent full-text assessment, which was always conducted independently by the same two reviewers (MP and MA), with the third reviewer (DF) assisting in case of disagreements. If the full-text of an eligible study could not be retrieved by the reviewers, the authors were contacted with a two-week reply window.

3.3.5 Data collection process and data items

Data extraction was conducted by one reviewer (MP) using a pretested extraction form that was specifically designed to align with the review's aims. A second reviewer (MA) verified the accuracy of the extracted data. If clarifications about the data were necessary (e.g., incomplete data), the authors were contacted with a two-week reply window. When data were available only in graphs or tables, WebPlotDigitizer (version 5.1) software was used to extract them (Drevon et al. 2017). Relevant data for each aspect of the PICOS framework were extracted. General study information (e.g., participant characteristics) and subjective measures of muscle fatigue (e.g., Borg scale ratings) were also reported. If studies included groups not relevant to the review's aims, they were not considered for data extraction.

3.3.6 Risk of bias assessment

Two independent reviewers (MP and MA) assessed the methodological quality of the included papers using the Appraisal tool for Cross-Sectional Studies (AXIS) (Downes et al. 2016). The original version of the tool consists of 20 questions divided into five different sections: introduction, methods, results, discussion, and other. However, the tool was slightly modified to better align with the objectives of this review, as previously reported (McArthur et al. 2020). Specifically, questions 7 and 14 were removed, as non-response bias had little relevance in our review due to the nature of the included studies. Question 13 of the original tool (now renumbered as question 12 in the modified version) was revised to assess whether information about participant dropouts during testing sessions were reported. Each question was answered with "Yes", "No", or "Don't know”. A response of “Yes” was awarded 1 point, while “No” or “Don’t Know” responses were awarded 0 points. However, question 19 of the original tool (now renumbered as question 17 in the modified version), which is related to the conflicts of interest, was scored differently. Due to the nature of the question, a response of “Yes” was awarded 0 points, while “No” received 1 point. The maximum possible score for each study was 18, with higher scores indicating lower risk of bias. To ensure comparability across studies, scores were converted into percentages. Studies scoring ≥ 75% were classified as “good” quality, those scoring between 50 and 74% as “moderate” quality, and those scoring <50% as “poor” quality (Lunt et al. 2021).

3.3.7 Synthesis methods

Both narrative and meta-analytic approaches were employed to synthesise the data from the included studies (Deeks et al. 2019; McKenzie et al. 2019). The narrative synthesis was conducted for endurance time, EMG parameters, and force decline while the meta-analysis was performed for endurance time only. No grouping was necessary for the meta-analysis, as all studies included the same outcome measure (endurance time in seconds), allowing for direct comparisons. The narrative synthesis involved outlining and tabulating study characteristics to facilitate a comprehensive interpretation of findings across all studies (McKenzie et al. 2019). For the meta-analysis, only studies with available or retrievable mean ± SD data on endurance times (in seconds) and sample sizes for both older and younger groups were included. The endurance results reported by Tsuboi et al. (2013) were pooled into a single group, as the authors originally reported data separately for male and female participants. The pooled mean and SD were calculated using the following equations:

$$\boldsymbol{Equation\ 1}.\ Mean\_pooled = \frac{(N\ male \times mean\_male) +\ (N\ female\ \times mean\_female)}{N\ male + N\ female}$$

$$\textbf{Equation 2}.\ SD\_pooled = \sqrt{\frac{[(N\ male - 1) \times\ SD^2\_male] + [(N\ female - 1) \times\ SD^2\_female]}{N\ male + N\ female - 2}}$$

where N represents the number of male/females in the respective group. The mean difference (MD) was selected as the outcome measure, as endurance time in seconds was a common metric across studies.

Meta-analysis procedures and forest plot generation were performed using R software (version 4.4.3) (R Core Team 2022) with the “meta” package (v. 8.0.2) (Balduzzi et al. 2019). A random effects model was chosen to account for the anticipated heterogeneity across studies, common in ageing research due to variations in study protocols, participant characteristics (including age ranges and fitness levels), sample sizes, task specificity, and measurement techniques. The Knapp-Hartung adjustment was applied to calculate the 95% confidence intervals around the pooled effect estimate (Jackson et al. 2017), providing more conservative estimates particularly suited for meta analyses with smaller numbers of studies. Additionally, prediction intervals were included alongside summary estimates to illustrate the expected range of true effects in future studies. This approach enhances clinical interpretation by showing the breadth of potential outcomes rather than just the average effect (Inthout et al. 2016; Deeks et al. 2019).

While a random-effects model was implemented to account for heterogeneity, it is important to acknowledge that this approach does not eliminate heterogeneity but accommodates it statistically (Deeks et al. 2019). Additional measures were taken to explore heterogeneity beyond the commonly used I^2 statistic, which has limitations in fully characterising heterogeneity patterns. To gain further insight into the variation in true effects across studies, the between-study variance (τ^2) was estimated using the restricted maximum-likelihood (REML) estimator, as recommended for continuous outcomes (Inthout et al. 2016; Deeks et al. 2019). Prediction intervals were also calculated for pooled effect sizes to offer a more comprehensive perspective on the potential range of true differences in endurance times between older and younger adults across studies (Inthout et al. 2016; Deeks et al. 2019). This approach enhances clinical interpretation by illustrating not only the average effect but also the expected range of effects in future studies.

3.3.8 Sensitivity analyses

Given the anticipated heterogeneity in ageing studies comparing endurance measures, comprehensive sensitivity analyses were conducted (Deeks et al. 2019). This involved sequentially excluding each study to assess the consistency of the meta-analysis results and evaluate the impact of individual studies on overall het erogeneity. Outliers and influential cases were also quantitatively explored using established deletion diagnostics adapted from linear regression (Viechtbauer 2010; Viechtbauer and Cheung 2010). These diagnostics included externally standardised residuals, DFFITS values, Cook's distances, covariance ratios, DFBETAS values, estimates of τ^2 and Q when each study was removed sequentially, diagonal elements of the hat matrix, and the weights assigned to observed outcomes during model fitting. These analyses were performed using the "influence" function from the "metafor" package (v. 4.6.0) in R (Viechtbauer 2010). The influence function allows for a quantitative identification of potential outliers and influential studies that may disproportionately affect the meta-analysis results, facilitating the detection of studies that could bias the findings. This comprehensive approach to identifying influential cases and conducting sensitivity analyses ensured the robustness of the findings regarding age-related differences in endurance times.

3.3.9 Certainty of evidence

The overall certainty of the evidence was independently assessed by two reviewers (MP and MA) using the Grading of Recommendations, Assessment, Development, and Evaluation (GRADE) approach. This involved a five-step process as out lined by Goldet and Howick (2013), resulting in a final rating of the quality of evidence as high, moderate, low, or very low. Based on GRADE guidelines, evidence from observational studies was initially rated as low quality, but could be upgraded in the presence of specific factors such as large effect sizes or clear dose-response relationships. Cohen's d values were used to estimate effect sizes. Conversely, the certainty was downgraded based on concerns related to risk of bias, inconsistency, imprecision, indirectness, or potential publication bias. Publication bias was assessed by generating funnel plots using the funnel function from the "metafor" package (v. 4.6.0) in R. Due to the small number of studies in our meta-analysis (fewer than 10), formal statistical tests such as Egger's regression test were not performed as they lack sufficient statistical power with small sample sizes and can lead to misleading results (Sterne and Egger 2001). Instead, we relied on visual inspection of the funnel plot to qualitatively evaluate potential asymmetry that might suggest publication bias. The GRADE approach was applied to the studies included in the meta-analysis, following the removal of the outlier (described below), to enable a tailored interpretation of the evidence in accordance with established guidelines for evaluating observational studies (Mueller et al. 2018).

3.4 Results

3.4.1 Study selection

The database search process yielded a total of 1253 records. Following duplicate removal, two independent reviewers screened the titles and abstracts of 1021 records. An additional paper was identified through hand-searching. Full-text screening was performed on 43 records in total (42 from databases and 1 from hand-searching). Four studies were excluded from this phase as the full text was not available: three were conference abstracts (Sagendorf et al. 2000; Ebenbichler et al. 2014; Piovanelli et al. 2019), and requests for the full text of one study were unsuccessful (Tsuboi et al. 2015). During the full-text screening, studies were excluded for the following reasons: 17 did not include the age groups of interest (i.e., older group < 60 years) (Latikka et al. 1995; Kankaanpaa et al. 1998; Valkeinen et al. 2002; Peolsson et al. 2007; Yassierli et al. 2007; Kell and Bhambhani 2008; Lariviere et al. 2009; Yassierli and Nussbaum 2009; van Dieën et al. 2009; Park et al. 2012; Kurz et al. 2014; Hiepe et al. 2015; Ebenbichler et al. 2017; Habenicht et al. 2020; Sibson et al. 2024; Vlazna et al. 2025; Chen et al. 2025); four lacked comparisons with younger groups (Gibbons et al. 1997; Laura Gibbons 1998; Ropponen et al. 2004; Keshavarzi et al. 2022); three involved individuals with low back pain or other medical conditions (Troup and Chapman 1972; Ebenbichler et al. 2020; Chen et al. 2023); one did not assess the muscles of interest (Lesniewski and Sinning 2000). The review ultimately included 14 studies (13 from databases and 1 from hand-searching). The complete screening process is illustrated in the PRISMA flow diagram in **Fig. 1** (Page et al. 2021).

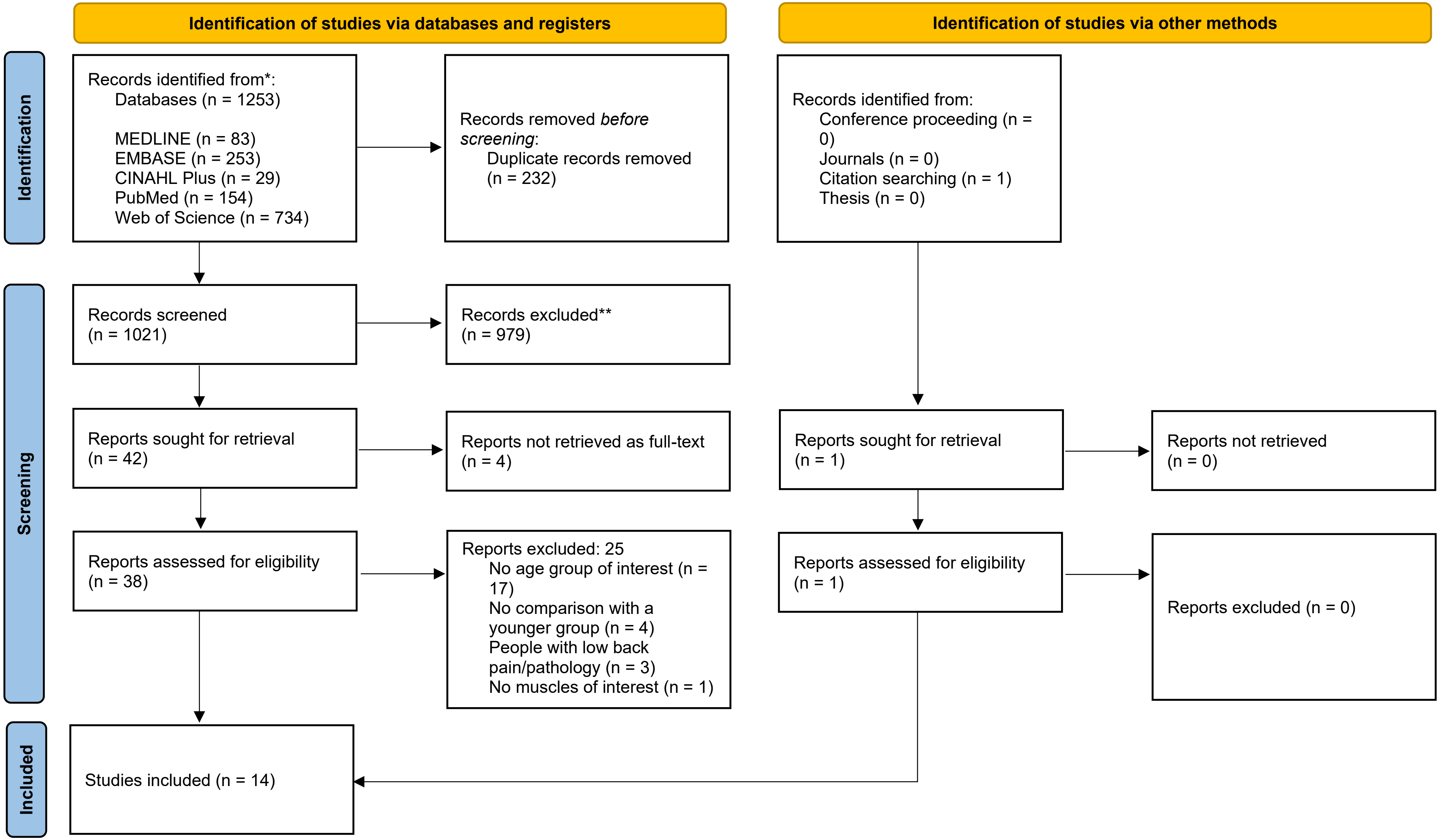

Fig. 1 PRISMA flow diagram.

3.4.2 Study characteristics

Although 14 studies were initially included in this review, the study by Parreira et al. (2014) was excluded due to suspected data overlap with a prior publication from the same authors (Parreira et al. 2013). This resulted in a final inclusion of 13 studies, the characteristics of which are summarised in **Table 1**. In total, these studies involved 1116 participants, of whom 326 were older individuals, while the remaining were young and middle-aged controls. All studies focused on fatigue/endurance of the back extensor muscles, while no studies investigating the neck extensor muscles in the context of fatigue and ageing were retrieved. When specific muscles were identified in the studies (i.e., in those assessing EMG), their names were explicitly reported in **Table 1** (e.g., multifidus, lumbar erector spinae). In contrast, when studies evaluated back muscles more generally, these muscles were described using the broader term “paraspinal muscles”. Additionally, all studies employed isometric tasks, except for Parreira et al. (2013), which used a dynamic fatiguing protocol.

3.4.3 Risk of bias assessment

A comprehensive summary of the risk of bias for each study is presented in **Table 2**. Scores ranged from 10 to 17 out of a maximum of 18, corresponding to percentages between 55.56% and 94.44%. None of the included studies has been rated as “poor” quality study. The full set of questions, along with the specific scores assigned to each question for each study, is presented in **Appendix - Supplemental file 3**.

Table 1 Characteristics of the 13 included studies.

Study	Elderly characteristics	Control group characteristics	Muscle/s assessed	Fatiguing task	Measurement tool	Outcome measures	Results
Koch et al. (2025)	**10 elderly** (65.0±5.0 years, 5 M, 5 F, 24.7±5.8 kg/m^2)	**10 young** (25.6±2.0 years, 5 M, 5 F, 22.3±2.3 kg/m^2)	**Paraspinal muscles**	**Modified Sørensen test on a 45° Roman Chair until voluntary exhaustion** (the trunk was maintained unsupported on the Roman Chair in a horizontal position relative to the ground)	/	**Endurance time**	**Endurance time results:** - <u>Elderly</u>: 222±98 s - <u>Young</u>: 368±154 s Mean ± SD values. *No statistical comparison was provided, as endurance time was a secondary outcome in the study. Therefore, only descriptive values are reported.
Parrella et al. (2025)	**13 elderly** (69±4 years, 8 M, 5 F, 172±8 cm, 70.23±8.17 kg, 23.58±1.80 kg/m^2) Mean ± SD values.	**13 young** (26±2 years, 8 M, 5 F, 173±13 cm, 70.62±15.19 kg, 23.32±2.44 kg/m^2) Mean ± SD values.	**Lumbar erector spinae muscle**	**Modified Sørensen test on a 45° Roman Chair at 30% maximal voluntary isometric force (MVIF) until voluntary exhaustion** (the trunk was maintained unsupported on the Roman Chair in a horizontal position relative to the ground)	**HD-sEMG placed on the right erector spinae** (64-electrode grid from L5 to L2 approximately)	**- Endurance time** **- Spatial distribution of muscle activity (x and y coordinates of the centroid of the topographical map)** at 0%, 25%, 50%, 75% and 100% of endurance time **- MFCV** at 0%, 25%, 50%, 75% and 100% of endurance time **- MVIF decline**	**No significant differences in endurance time** between the two age groups (p=0.749): - <u>Elderly</u>: 99.74±34.03 s - <u>Young</u>: 104.32±38.17 s **Significant age-related differences for centroid displacement along both the x** (F = 6.043, η_p^2 = 0.201, p = 0.006) **and y axis** (F = 5.111, η_p^2 = 0.176, p = 0.017) **over time**: - <u>Elderly</u>: no significant changes along the x axis (p>0.05), but cranial shift of the centroid from 0% to 25% (p=0.003), 0% to 75% (p = 0.003), 0% to 100% (p = 0.004), 50% to 75% (p=0.002), and from 50% to 100%

						(%) after the task	(p=0.006); - Young: lateral displacement of the centroid between 0% and 100% (p = 0.004), but no changes along the y-axis (p>0.05). **No differences in muscle fibre conduction velocity decline between groups** (F = 0.177, η_p^2 = 0.007, p = 0.752). **No differences in MVIF decline between the two groups** (p=0.681): - Elderly: −22.70±17.67% - Young: −20.45±8.2% Mean ± SD values.
Han et al. (2024)	**59 elderly** (65.95±4.48 years, 29 M, 30 F, 164.71±8.65 cm, 66.44±9.22 kg, 24.46±2.53 kg/m^2); - **Physical activity index (PAI)**: 25.52±8.11 Mean ± SD values.	**29 young** (27.41±4.32 years, 14 M, 15 F, 169.69±7.79 cm, 66±15.34 kg, 22.65±3.59 kg/m^2); - **PAI**: 39.58±5.45; **93 middle-aged** (51.85±4.72 years, 43 M, 50 F, 164.23±8.69 cm, 68.11±13.69 kg, 25.17±4.41 kg/m^2); - smoking (yes):	**Paraspinal muscles**	**Ito test** performed until task failure or for a maximum of 5 min (prone position on examination table with a pad under the lower abdomen; trunk raised ~15° with arms parallel to the body, cervical spine neutral, and feet maintained on the table)	/	**Endurance time**	**↓ Endurance time in older adults** compared to both young and middle-aged subjects (p<0.001): - Elderly: 120.07±78 s - Middle-aged: 162.22±73.77 s - Young: 182.97±51.78 s **↓ Physical activity level in elderly** compared to both young and middle-aged subjects (p<0.001) Mean ± SD values.

		22; - **PAI**: 30.38±6.65 Mean ± SD values.					
Liu et al. (2024)	**18 elderly** (65.17±4.72 years, 24.08±1.92 kg/m^2) Mean ± SD values.	**27 young** (29.44±5.71 years, 21.00±2.98 kg/m^2); **49 middle-aged** (49.73±4.57 years, 22.76±2.11 kg/m^2) Mean ± SD values.	**Paraspinal muscles**	**Ito test** performed until task failure (prone position on examination table with a pad under the lower abdomen; trunk raised ~15° with arms parallel to the body, cervical spine neutral, and feet maintained on the table)	**/**	**Endurance time**	**No significant differences in endurance time** between the three groups (p=0.135): - Elderly: 36.22±30.69 s - Middle-aged: 53.74±43.99 s - Young: 63.72±46.04 s Mean ± SD values.
Da Silva et al. (2015)	**10 elderly** (73±7 years, 160±10 cm, 67±10 kg, 26±4 kg/m^2). Mean ± SD values.	**10 young** (30±4 years, 165±12 cm, 64±8 kg, 23±2 kg/m^2). Mean ± SD values.	**Multifidus** and **iliocostalis lumborum muscles**	2 different isometric fatiguing protocols: - **45° Roman chair exercise** with the trunk maintained unsupported in a horizontal position relative to the ground **for 60 s;** - **Functional task** consisting of maintaining a box, equivalent to 10% of body mass, close to the trunk in the upright position **for 60 sec.**	**8 sEMG bipolar electrodes placed bilaterally on the multifidus** at L5 level and **iliocostalis** at L3 level	**- Normalised EMG index of muscle fatigue** (NMFslp) defined as the decline in the MF over time - **BORG CR-10 scale** throughout the fatiguing tasks	**No significant differences in NMFslp** and **BORG scale values** between age groups during the **Roman chair exercise**: - Multifidus (p=0.143): -0.17±0.18 %/s in elderly, -0.17±0.23 %/s in young - Iliocostalis (p=0.079): -0.11±0.16 in elderly, -0.25±0.15 %/s in young - Borg (p=0.206) **No significant differences in NMFslp** and **BORG scale values** between age groups during the **functional task**:

							- <u>Multifidus</u> (p=0.689): -0.10±0.14 %/s in elderly, -0.12±0.29 %/s in young - <u>Iliocostalis</u> (p=0.217): -0.10±0.23 %/s in elderly, -0.20±0.24 %/s in young - <u>Borg</u> (p=0.380) Results from each muscle are averaged between the two side (mean ± SD values).
Kienbacher et al. (2014)	**42 elderly** (21 F, 67.11±1.55 years, 25.15±0.45 kg/m^2) - **IPAQ score**: 424.09±59.07. Mean ± SE values.	**44 young** (19 F, 33.14±1.66 years, 24.11±0.47 kg/m^2) - **IPAQ score**: 337.15±60.25 Mean ± SE values.	**Multifidus, longissimus and iliocostalis lumborum muscles**	**Sustained isometric back extension test at 80% of MVC for 30s** (participants were seated on the device with the trunk flexed forward at 30°)	**6 double parallel-bar sEMG electrodes placed bilaterally on the multifidus** at L5 level, **longissimus** at L2 level and **iliocostalis** at L1 level	**Absolute and normalised MF slope declines** over the task	**↓ Absolute MF slope decline in elderly at L5 level only** (-0.14±0.03 Hz/s in elderly, -0.24±0.04 in young; F=04.26, p=0.04) **↓ Normalised MF slope decline in elderly at L5 level only** (-0.12±0.03 %/s in elderly, -0.26±0.04 %/s in young; F=06.40, p=0.01) **No differences in physical activity level between age groups** (p>0.05) Mean ± SE values.
Tsuboi et al. (2013)	**- 11 elderly men** (71.1±3.6 years, 168±4.8 cm, 67.0±7.1 kg, 23.8±2.1 kg/m^2)	**- 17 young men** (25.1±3.2 years, 172±3.9 cm, 63.2±5.5 kg, 21.4±1.8 kg/m^2)	**Lumbar erector spinae muscle**	**Biering-Sørensen back endurance test until voluntary exhaustion** (prone on a bed with the anterior superior	**One active sEMG electrode on the left erector spinae** at L1 level	**- Slope of MF values** and **MPF values** over the task **- Endurance**	**Elderly men:** **- ↓ MF slopes** compared to young men (-21.6±11.8 % per min in elderly, -29.3±8.1 % per min in young;

	- 11 elderly women (67.6±3.3 years, 153±3.7 cm, 53.9±7.1 kg, 22.9±3.0 kg/m^2). Mean ± SD values.	**- 14 young women** (25.6±1.8 years, 158±5.3 cm, 51.5±4.7kg, 20.7±1.3 kg/m^2). Mean ± SD values.		iliac spines at the edge; trunk maintained in a horizontal position relative to the lower body with hands on the head)	(with one reference electrode on L1 spinous process)	**time**	p<0.05) **- ↓ MPF slopes** compared to young men (-15.6±6.4 % per min in elderly, -24.5±7.4 % per min in young; p<0.01) **- no differences in endurance time** (95±44 s in elderly, 98±21 s in young; p>0.05) **Elderly women:** **- no differences in MF slopes** compared to young women (-17.1±7.9 % per min in elderly, 19.9±5.5 % per min in young; p>0.05) **- no differences in MPF slopes** compared to young women (-12.8±4.9 % per min in elderly, 11.1±4.5 % per min in young; p>0.05) **- no differences in endurance time** compared to young women (129±59 s in elderly, 143±48 s in young; p>0.05) Mean ± SD values.
Parreira et al. (2013)	**18 elderly** (69.8±4.9 years, 9 M, 157±7 cm, 65.9±9 kg, 26.5±3.1 kg/m^2) Mean ± SD values.	**18 young** (26.3±5.8 years, 9 M, 168±10 cm, 64.6±13.3 kg, 22.4±2.6 kg/m^2) Mean ± SD values.	**Paraspinal muscles**	**Dynamic back extension exercise on a 45° Roman Chair until voluntary exhaustion** (2 s of flexion and 2 s of extension from 45° to 0° relative to the ground at 60/bpm)	**/**	**Endurance time**	**↑ Endurance time in elderly** compared to young (p=0.039): - Elderly: 133±52 s - Young: 97±27 s Mean ± SD values.

Bašič et al. (2013)	**14 elderly men** (72.0±7.2 years, 172±6 cm, 77.9±13.5 kg). Mean ± SD values.	**16 young men** (27.5±4.1 years, 178±5 cm, 74.5±8.1 kg). Mean ± SD values.	**Paraspinal muscles**	**Modified Sørensen test on a 45° Roman Chair until voluntary exhaustion** (the trunk was maintained unsupported on the Roman Chair in a horizontal position relative to the ground)	**/**	**Endurance time**	**No differences in endurance time** between elderly and young participants (p=0.82): - Elderly: 3.1±1.8 min - Young: 4.2±1.2 min Mean ± SD values.
Adedoyin et al. (2011)	**Group aged 60+ (*N=31*)** (62.8±3.0 years, 165±10 cm, 69.2±11.7 kg, 25.1±4.4 kg/m^2). Mean ± SD values.	**- Group aged 19-29 (*N=210*)** (23.2±2.4 years, 168±10 cm, 63.3±9.7 kg, 22.4±3.2 kg/m^2). **- Group aged 30-39 (*N=103*)** (33.4±2.9 years, 166±10 cm, 64.5±11.7 kg, 23.3±4.1 kg/m^2). **- Group aged 40-49 (*N=128*)** (43.4±3.0 years, 165±10 cm, 66.9±13.7 kg, 24.2±4.5 kg/m^2). **- Group aged 50-59 (*N=89*)**	**Paraspinal muscles**	**Biering-Sørensen back endurance test until voluntary exhaustion** (prone on a bed with the anterior superior iliac spines at the edge; trunk maintained in a horizontal position relative to the lower body with arms parallel to the body)	**/**	**Endurance time**	**↓ Endurance time with increasing age** (F=32.702; p=0.001) - Group aged 19-29: 133±41 s - Group aged 30-39: 121±49 s - Group aged 40-49: 103±42 s - Group aged 50-59: 82±36 s - Group aged 60+: 81±33 s Endurance time differed significantly between paired groups (p<0.05), except between age groups 50–59 and 60+years (p>0.05). Mean ± SD values.

		(53.8±2.6 years, 164±10 cm, 66.7±12.2 kg, 24.8±4.6 kg/m²). Mean ± SD values.					
Mbada et al. (2011)	**Group aged 60-69 (*N=47*)** (62.0±2.56 years, 165±7 cm, 67.9±11.4 kg, 24.9±4.13 kg/m²). Mean ± SD values.	**- Group aged 20-29 (*N=336*)** (23.7±2.32 years, 167±9 cm, 62.6±10.2 kg, 22.4±3.42 kg/m²). **- Group aged 30-39 (*N=143*)** (33.3±2.92 years, 166±9 cm, 63.3±11.0 kg, 23.0±3.94 kg/m²). **- Group aged 40-49 (*N=199*)** (43.4±3.05 years, 165±9 cm, 66.6±13.2 kg, 24.4±4.61 kg/m²). **- Group aged 50-59 (*N=199*)** (53.8±2.55 years, 164±9 cm, 66.6±13.2 kg, 24.7±4.71 kg/m²). Mean ± SD values.	**Paraspinal muscles**	**Biering-Sørensen back endurance test until voluntary exhaustion** (prone on a bed with the anterior superior iliac spines at the edge; trunk maintained in a horizontal position relative to the lower body with arms parallel to the body)	**/**	**Endurance time**	**↓ Endurance time with increasing age** (F=54.197; p=0.001) - Group aged 11-19: 137.0±64.8 s - Group aged 20-29: 138.4±42.6 s - Group aged 30-39: 121.9±47.8 s - Group aged 40-49: 102.7±41.2 s - Group aged 50-59: 81.8±36.2 s - Group aged 60-69: 77.9±31.3 s Endurance time differed significantly between paired groups (p<0.05). Mean ± SD values.

Singh et al. (2011)	**26 elderly** (72.1±5.9 years, 10 M, 16 F). Mean ± SD values.	**26 young** (27.9±5.2 years, 10 M, 16 F). Mean ± SD values.	**Longissimus muscle**	**Isometric lumbar extension contraction at 60% MVC for 120 s** (participants stood upright facing a load cell connected at the 12th thoracic level by a metal cable in a restraining frame with the pelvis and legs immobilised)	**2 bipolar EMG electrodes placed bilaterally on the longissimus muscle** at L3 level	**Power of the EMG signals** in 5 different frequency bands (20-100 Hz; 101-200 Hz; 201-300 Hz; 301-400 Hz and 401-499 Hz) at 1 s, 40 s, 80 s and 120 s of task duration	**↑ Power of the EMG signals in the 101-200 Hz band in elderly** compared with younger adults in all time epochs ($p<0.05$): - <u>1 s</u>: 29.46±4.23 % in elderly, 25±5.10 % in young - <u>40 s</u>: 29.82±4.09 % in elderly, 25.44± 4.45 % in young - <u>80 s</u>: 29.16±3.8 % in elderly, 25.15± 4.31 % in young - <u>120 s</u>: 29.89±3.22 % in elderly, 25±4.16 % in young Results are averaged between the two sides. Mean ± SD values.
Champagne et al. (2009)	**16 elderly men** (72.8±4.7 years, 171±10 cm, 79±6.7 kg, 27.1±2.6 kg/m^2). Mean ± SD values.	**20 young men** (22.8±3.1 years, 1.79±10 cm, 77.3±10.8 kg, 24.1±2.5 kg/m^2). Mean ± SD values.	**Paraspinal muscles**	**Modified Sørensen test on a 45° Roman Chair until voluntary exhaustion** (the trunk was maintained unsupported on the Roman Chair in a horizontal position relative to the ground)	/	**- Endurance time** **- Decline in maximal voluntary isometric lift force (%)** after the fatiguing task **- Borg CR-10 scale** measured throughout the task at successive 10% intervals of total endurance time - **Time to**	**No differences in endurance time between groups** (296.3±84.1 s in elderly, 334.9±80.4 s in young; $p=0.17$) **No differences in maximal isometric lift force decline between groups** (10.3%±9.0% in elderly; 5.9%±9.2% in young; $T=-1.433$; $p=0.16$) **No between-group differences in Borg scale** throughout the fatiguing task ($F=2.62$; $p=0.11$) - <u>10%</u>: 1.7±0.26 in elderly,

						exhaustion at Borg-7 rate (expressed as % of total endurance time)	1.51±0.23 in young - 20%: 2.26±0.27 in elderly, 2.07±0.29 in young - 30%: 3.26±0.26 in elderly, 2.87±0.27 in young - 40%: 4.62±0.36 in elderly, 3.76±0.2 in young - 50%: 5.85±0.37 in elderly, 4.89±0.35 in young - 60%: 7.03±0.44 in elderly, 6.21±0.35 in young - 70%: 8.04±0.39 in elderly, 7.33±0.34 in young - 80%: 9.02±0.28 in elderly, 8.54±0.25 in young - 90%: 9.64±0.22 in elderly, 9.31±0.15 in young - 100%: 10 in elderly, 10 in young **No differences between age groups in the time remaining to exhaustion at Borg-7 rate** (43.40±3.1 % in elderly; 38.58±2.61 % in young; T=-1.22; p=0.23) Mean ± SEM values for force decline, Borg and time at Borg-7 rate. Mean ± SD values for endurance time.

= good quality (≥75%)
= moderate quality (50-74%)
= poor quality (<50%)

Study	Introduction	Methods	Results	Discussion	Other	SCORE	%
Koch et al. (2025)	1/1	9/9	3/4	2/2	2/2	17/18	94.44
Parrella et al. (2025)	1/1	8/9	3/4	2/2	2/2	16/18	88.89
Han et al. (2024)	1/1	8/9	2/4	2/2	2/2	15/18	83.33
Liu et al. (2024)	1/1	7/9	1/4	1/2	2/2	12/18	66.67
Da Silva et al. (2015)	1/1	7/9	2/4	2/2	2/2	14/18	77.78
Kienbacher et al. (2014)	1/1	5/8	3/4	2/2	2/2	13/18	72.22
Parreira et al. (2013)	1/1	7/9	2/4	2/2	2/2	14/18	77.78
Tsuboi et al. (2013)	1/1	6/9	2/4	1/2	1/2	11/18	61.11
Bašič et al. (2013)	1/1	6/9	2/4	0/0	1/2	10/18	55.56
Adedoyin et al. (2011)	1/1	7/9	2/4	2/2	1/2	13/18	72.22
Mbada et al. (2011)	1/1	7/9	2/4	2/2	1/2	13/18	72.22
Singh et al. (2011)	1/1	8/9	1/4	1/2	1/2	12/18	66.67
Champagne et al. (2009)	1/1	7/9	4/4	2/2	2/2	16/18	88.89

Table 2 Risk of bias assessment of the included studies using the Appraisal tool for Cross-Sectional Studies (AXIS).

3.4.4 Narrative synthesis of the results

The main findings from the individual studies were grouped per outcome domains and described in a narrative way. Ten studies assessed endurance time (Champagne et al. 2009; Adedoyin et al. 2011; Mbada et al. 2011; Bašič et al. 2013; Parreira et al. 2013; Tsuboi et al. 2013; Han et al. 2024; Liu et al. 2024; Parrella et al. 2025; Koch et al. 2025), five studies evaluated EMG parameters (Singh et al. 2011; Tsuboi et al. 2013; Kienbacher et al. 2014; da Silva et al. 2015; Parrella et al. 2025) and two studies investigated force decline (Champagne et al. 2009; Parrella et al. 2025).

3.4.4a Endurance time

Three studies reported reduced endurance time in older compared to younger individuals, specifically during the Biering-Sørensen endurance test (Adedoyin et al. 2011; Mbada et al. 2011) and the Ito test (Han et al. 2024). In contrast, five studies found no significant differences in endurance time between younger and older participants (Champagne et al. 2009; Bašič et al. 2013; Tsuboi et al. 2013; Liu et al. 2024; Parrella et al. 2025). Three of these studies employed a modified version of the Sørensen test performed on a 45° Roman Chair until voluntary exhaustion (Champagne et al. 2009; Bašič et al. 2013; Parrella et al. 2025), with Parrella et al. (2025) using a submaximal intensity set at 30% of maximal voluntary isometric force (MVIF). Instead, the other two studies used the Ito test (Liu et al. 2024) and the standard Biering-Sørensen test (Tsuboi et al. 2013). In the study by Koch et al. (2025), which employed a modified version of the Sørensen test, endurance time was reported only as a secondary outcome, and no statistical comparison between age groups was conducted. Therefore, no specific findings are discussed in the narrative synthesis; however, the descriptive data (**Table 1**) were included in the meta-analysis. Lastly, one paper reported greater endurance time in older individuals compared to younger controls during a dynamic back extension task until voluntary exhaustion (Parreira et al. 2013).

3.4.4b EMG parameters

Three studies assessed the decline in MF slope during fatiguing tasks for the back extensors (Tsuboi et al. 2013; Kienbacher et al. 2014; da Silva et al. 2015), with Tsuboi et al. (2013) also evaluating the decline in MPF. One study found a reduced decline in both absolute and normalised MF values in older adults compared to younger controls for the erector spinae assessed at the L5 level during a 30-s isometric back extension task at 80% MVC (Kienbacher et al. 2014). Similarly, Tsuboi et al. (2013) reported lower MF and MPF slope declines in the lumbar erector spinae of older men during the Biering-Sørensen test performed until voluntary exhaustion, although no such differences were observed in older women. Additionally, no differences in MF decline between age groups were observed in the multifidus and iliocostalis lumborum muscles by da Silva et al. (2015) during either a 60-s 45° Roman chair exercise or a functional task involving holding a box in an upright position for 60 s. In contrast, Singh et al. (2011) found that older adults exhibited higher EMG signal power in the 101-200 Hz frequency band of the longissimus muscle compared to younger adults during all time epochs of a 120-s isometric lumbar extension task performed at 60% MVC. Lastly, Parrella et al. (2025), using high-density electromyography (HD-sEMG), found comparable declines in MFCV of the lumbar erector spinae muscle between age groups during a modified version of the Biering-Sørensen test performed at 30% MVIF to exhaustion. However, age-related differences in the spatial distribution of muscle activity were observed, with older adults exhibiting a cranial shift in the centroid of muscle activity with fatigue, while younger individuals showed a lateral displacement.

3.4.4c Force decline

One study assessed the decline in maximal isometric lift force following a modified version of the Sørensen test performed until voluntary exhaustion (Champagne et al. 2009) and found no differences between older and younger individuals. Similarly, Parrella et al. (2025) observed a com parable reduction in MVIF across age groups using the same test, but performed at a submaximal intensity (30% MVIF).

3.4.5 Meta-analysis results for endurance time

3.4.5a Initial analysis including all studies

A random-effects meta-analysis was initially conducted including all 10 identified studies, comprising 958 participants (248 elderly and 710 young adults). This preliminary analysis indicated that the elderly exhibited significantly reduced endurance of the back extensor muscles compared to younger adults (MD = −33.91 s [95% CI: −62.55; −5.26], t9 = −2.68, p = 0.025; **Appendix - Supplemental file 4**). However, substantial heterogeneity was observed (I^2 = 87.0%, τ^2 = 1091.87, χ^2 = 69.27, $p < 0.001$), warranting further investigation of studies potentially contributing to this heterogeneity.

3.4.5b Identification of outliers

Influence diagnostics were performed to identify potential outliers that might disproportionately affect the meta-analysis results. This quantitative assessment clearly identified Parreira et al. (2013) (study 5) as a significant outlier (**Appendix - Supplemental file 5**). This finding was consistent across several diagnostic measures, particularly evident in the Studentised Residuals, DFFITS values, Cook's Distances, Covariance ratios, estimates of heterogeneity (τ^2 and Q-statistic), and DFBETAS plots. The specific values from the influence diagnostics analysis are provided in **Appendix - Supplemental file 6**. The methodological assessment confirmed this statistical identification, as Parreira et al. (2013) was the only study assessing trunk fatigue during a dynamic task, whereas all other studies evaluated trunk muscle endurance during a sustained back extension isometric contraction. Additionally, it was the only study showing significantly better endurance in older adults compared to younger adults (MD = 36.00 [95% CI: 8.93; 63.07]), contrary to the pattern observed across all other included studies. Due to these statistical and methodological considerations, we determined that Parreira et al. (2013) should be excluded from the primary analysis.

3.4.5c Refined analysis after outlier removal

After excluding Parreira et al. (2013), our refined meta-analysis included 9 studies with a total of 922 participants (230 elderly and 692 young adults). The results demonstrated a larger and more statistically significant effect, with older adults exhibiting markedly reduced endurance times of the back extensor muscles compared to younger adults (MD = −41.31 [95% CI: −64.04; −18.57], $t_8 = -4.19$, $p = 0.003$; **Fig. 2**). Heterogeneity, although still present, was substantially reduced ($I^2 = 76.4\%$, $\tau^2 = 485.07$).

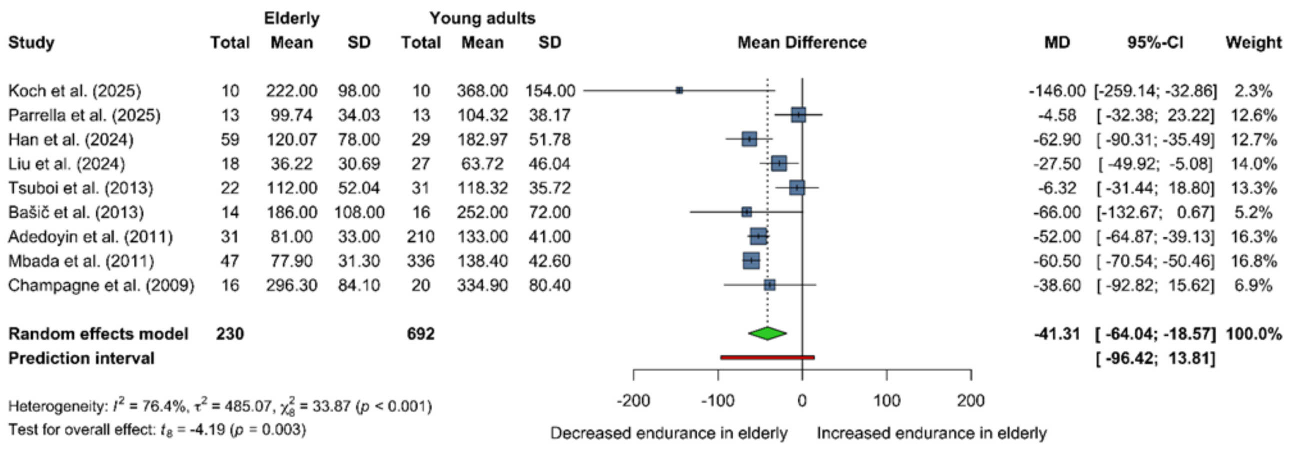

Fig. 2 Forest plot of endurance time differences between elderly and young adults after outlier removal. The plot of the refined analysis displays mean differences (MD) in endurance time (seconds) with corresponding 95% confidence intervals (95% CI) for each included study. The green diamond represents the pooled effect estimate with its 95% CI, while the red line indicates the prediction interval. Negative values on the x-axis indicate decreased endurance time in elderly compared to young adults, while positive values indicate increased endurance in elderly. The size of each square is proportional to the study's weight in the meta-analysis. The forest plot is organised in descending order based on publication year.

Secondary influence diagnostics performed on the refined dataset did not identify any studies as potential outliers (**Fig. 3**). The specific values from the influence diagnostics analysis are provided in **Appendix - Supplemental file 7**. Sensitivity analyses confirmed the robustness of our findings, as the effect sizes remained significant across all iterations regardless of which study was removed from the model (**Fig. 4**).

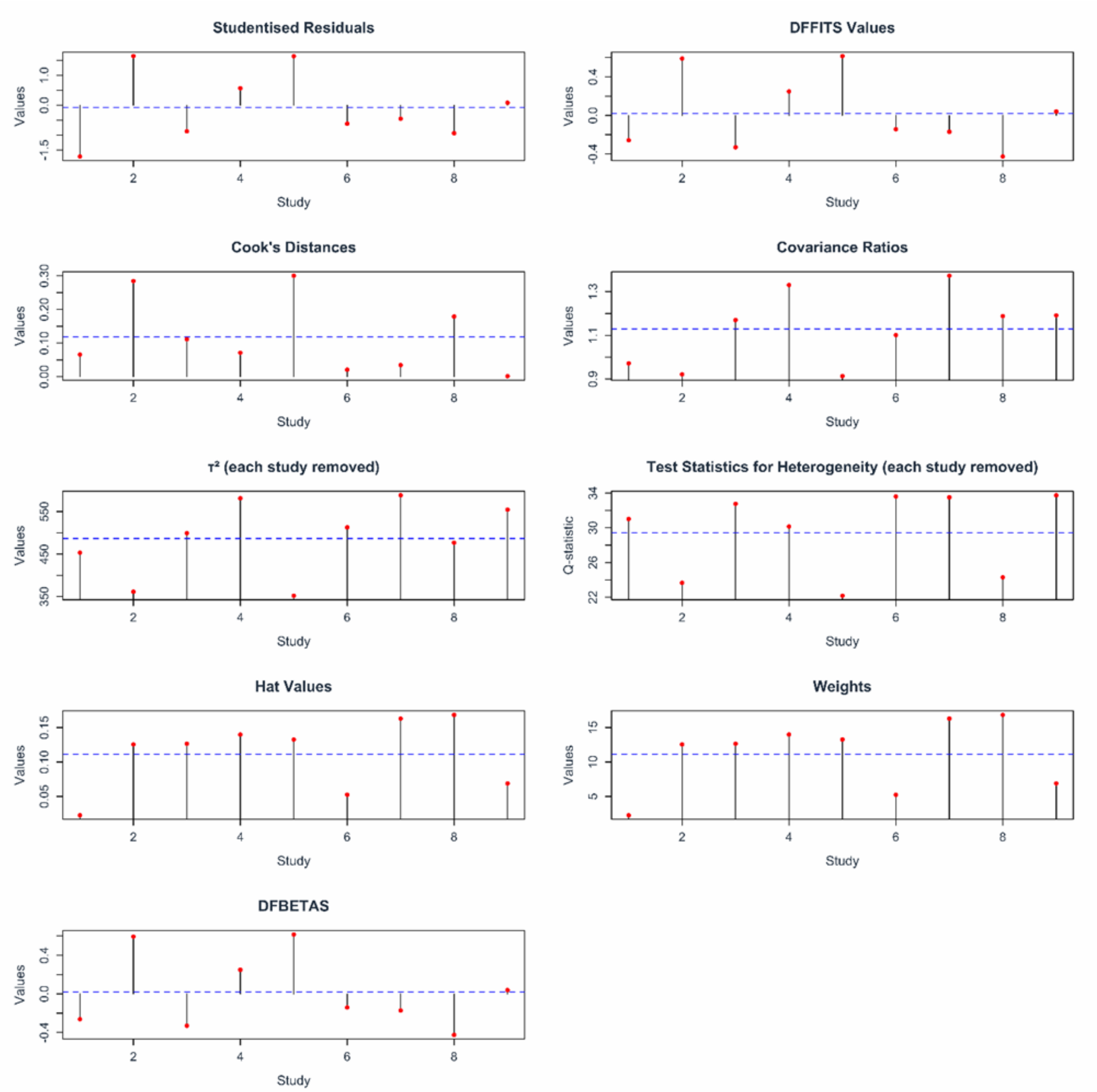

Fig. 3 Influence diagnostic plots for the meta-analysis of endurance time differences between elderly and young adults. The figure presents an influence analysis for the meta-analytic model, illustrating various diagnostic measures to identify influential studies. The plots include externally studentised residuals, DFFITS values, Cook's distances, covariance ratios, estimates of heterogeneity (τ^2 & Q-statistic) when each study is removed, hat values, weights and DFBETAS. The y-axis shows the values of each measure, and the x-axis displays the study numbers, ordered according to their appearance in the forest plot. The blue line in each plot represents the mean of each parameter across all studies.

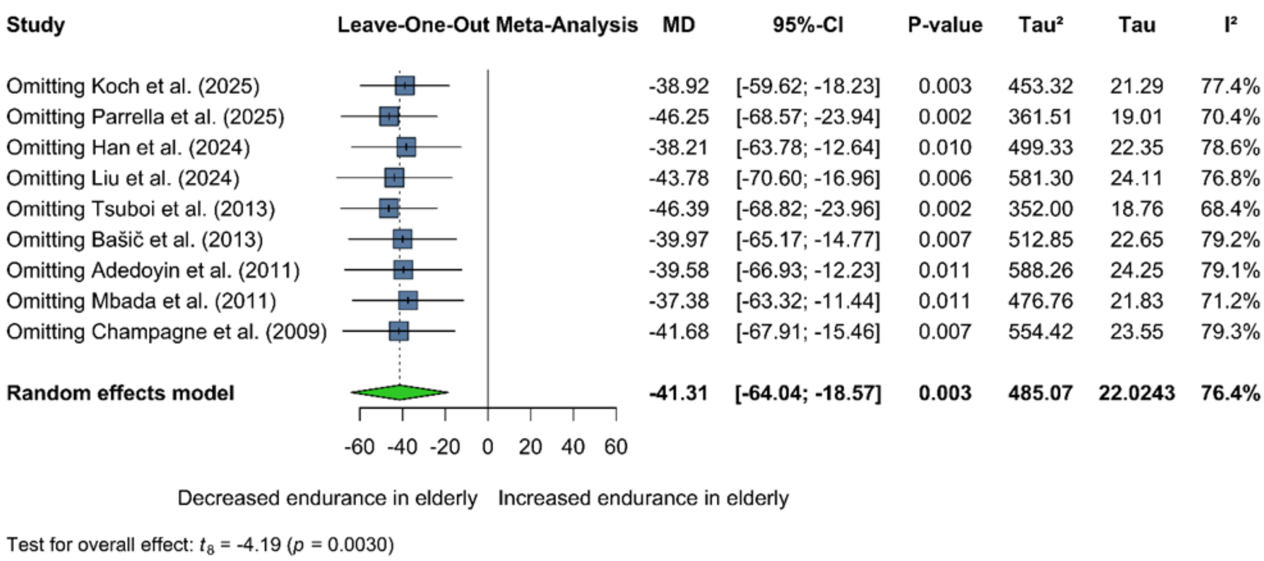

Fig. 4 Forest plot showing sensitivity analysis with sequential removal of individual studies after exclusion of Parreira et al. (2013). Mean differences (MD) with 95% confidence intervals are shown for the meta-analysis after sequentially omitting each remaining study, following the exclusion of Parreira et al. (2013). The green diamond at the bottom represents the overall effect from the analysis. Each row shows how the pooled effect estimates, confidence intervals, and heterogeneity statistics (P-value, Tau², Tau, and I²) change when the corresponding study is removed.

3.4.5d Assessment of publication bias

The funnel plot (**Fig. 5**) shows a relatively symmetrical distribution of studies around the central effect estimate, with studies appearing on both sides of the mean effect. While there is some clustering of studies on the left side (negative effects), this is consistent with the overall negative effect observed in the meta-analysis rather than indicative of bias. The reason able symmetry of the plot suggests that publication bias is unlikely to substantially influence our findings, although caution is warranted given the small number of included studies.

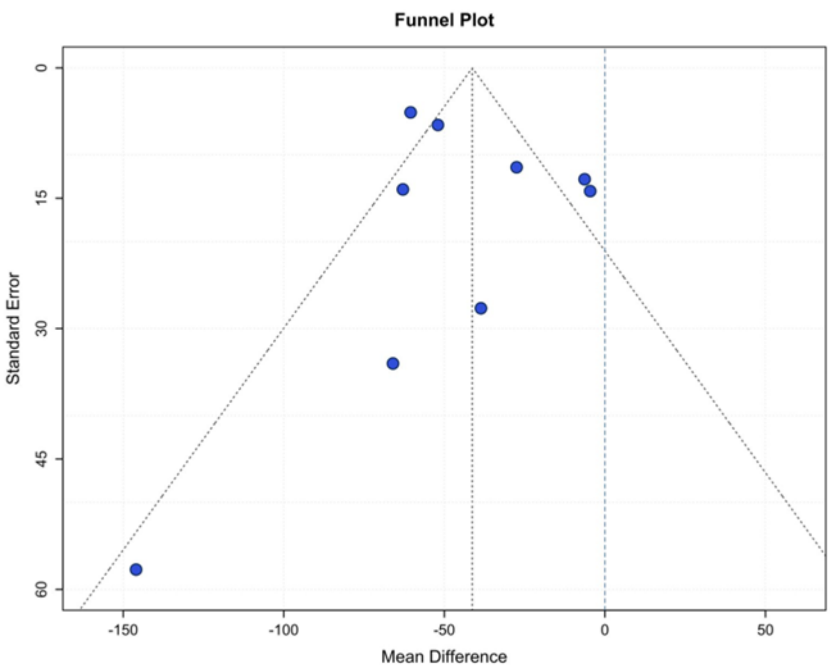

Fig. 5 Funnel plot for the assessment of publication bias. The blue dots represent individual studies, with mean difference (seconds) on the x-axis and standard error on the y-axis. The vertical black dashed line indicates the overall effect.

3.4.5e GRADE

Considering all the domains assessed using the GRADE approach, the findings from the meta-analysis provide moderate certainty of evidence that the endurance time of the back extensor muscles during isometric tasks is reduced with ageing. The included studies were not downgraded for risk of bias or publication bias as the AXIS tool and the funnel plot did not reveal any major concerns, respectively. For inconsistency, no downgrade was needed despite high heterogeneity (I^2 = 76.4%) because the direction of the effect was consistent across studies. Therefore, the substantial statistical heterogeneity likely reflects methodological differences between studies, such as variations in sample size, rather than true inconsistency in the effect estimates. Additionally, indirectness was not a concern given that the outcome variable (endurance time) was consistently measured across all studies, and the target population was appropriately represented. No downgrade was applied for imprecision because the confidence interval did not cross the line of no effect (zero), with both upper and lower bounds indicating a meaningful difference. Lastly, an upgrade was applied as the calculated Cohen's d for our outcome indicated a medium to large effect size (-0.77). The summary of the results is presented in **Table 3**.

Study	Outcome	Rating	Reasons for downgrading (↓) or upgrading (↑)
Koch et al. (2025) Parrella et al. (2025) Han et al. (2024) Liu et al. (2024) Tsuboi et al. (2013) Bašič et al. (2013) Adedoyin et al. (2011) Mbada et al. (2011) Champagne et al. (2009)	Reduced endurance with ageing during isometric back extension tasks	○○⊕○ MODERATE certainty of evidence	↑ Medium to large Cohen's d (-0.77)

Table 3 Summary of the overall certainty of evidence.

3.5 Discussion

The aim of this systematic review and meta-analysis was to synthesise the existing evidence on how ageing influences the development of muscle fatigue in the spinal extensor muscles, including the lumbar, thoracic, and cervical (neck) extensors. However, only findings related to the back extensor muscles are reported and discussed, as no studies were identified that examined and compared fatigue in the neck extensors between younger and older adults. Based on data from 9 studies, a meta-analysis revealed, with moderate certainty of evidence, a significant reduction in endurance time of the back extensor muscles during sustained isometric con tractions in older adults. In contrast, EMG studies investigating neuromuscular fatigue in the back extensor muscles produced mixed findings, likely due to considerable methodological variability across studies.

Is endurance of the back extensor muscles reduced in older adults?

Across all studies included in the meta-analysis, the observed effect consistently suggested a decline in endurance time with ageing (**Fig. 2**). The substantial heterogeneity among studies is likely attributable to differences in participant characteristics (e.g., sample size). Additionally, there may be an overlap in the populations reported by two studies (Adedoyin et al. 2011; Mbada et al. 2011), as their characteristics appear very similar, but one study involves a larger sample size. Attempts to clarify this potential overlap with the authors were unsuccessful. Nevertheless, sensitivity analysis confirmed the robustness of the findings, as the exclusion of any single study did not alter the overall outcome.

Contrary to our findings, a previous systematic review by Christie et al. (2011) reported a general increased fatigue resistance in elderly people compared to young adults during isometric contractions of lower and upper limb muscles. This phenomenon, defined as the "fatigue paradox", has been attributed to physiological changes associated with ageing, such as greater loss and atrophy of fast-twitch muscle fibres, motor unit remodelling, and a reduced reliance on glycolytic metabolism (Hunter et al. 2016). Altogether, these factors contribute to slower contractile properties in older adults. However, this mechanism may not apply as prominently to the back extensor muscles (e.g., the erector spinae), which already exhibit a high proportion of slow-twitch fibres in younger individuals (approximately 60–70%) (Jørgensen et al. 1993). Therefore, although age-related physiological changes likely occur in these muscles as well, the functional consequences may not be as evident as in muscles with a greater proportion of fast-twitch fibres, such as the elbow flexors and knee extensors. Nevertheless, in addition to muscle-specific properties, the type of isometric tasks used to assess back extensor endurance in the present meta-analysis may also be relevant. Specifically, the tasks included the standard Biering-Sørensen test (Adedoyin et al. 2011; Mbada et al. 2011; Tsuboi et al. 2013), a modified version of the Biering-Sørensen test using a 45° Roman chair (Champagne et al. 2009; Bašič et al. 2013;

Parrella et al. 2025; Koch et al. 2025), and the Ito test (Han et al. 2024; Liu et al. 2024). These tasks are more complex than the isolated, joint-specific protocols commonly used to assess endurance in upper and lower limb muscles. Indeed, these widely used protocols for assessing back muscle function across diverse populations (Demoulin et al. 2006) are multi-joint postural tasks that require coordinated activation not only of the trunk muscles but also of the hip extensors, together with contributions from stabilising muscles and passive structures. Such differences in task characteristics, in addition to differences in muscle properties, may influence the manifestation of age-related fatigue.

Moreover, Liu et al. (2024) observed a positive correlation between the functional cross-sectional area (FCSA) of the multifidus muscle and static back muscle endurance. In contrast, other studies did not report any significant associations between the CSA of the paraspinal muscles and endurance time (Gibbons et al. 1997; Yazici and Yerlikaya 2022; Han et al. 2024). However, there is consistent evidence indicating a negative correlation between fat infiltration within the paraspinal muscles and back endurance (Yazici and Yerlikaya 2022; Han et al. 2024; Liu et al. 2024). Muscle quality (e.g., fat infiltration) is a key determinant of muscle function (Dallaway et al., 2020), particularly in the paraspinal muscles, which are predominantly composed of slow-twitch fibres that tend to accumulate greater amounts of intramyocellular lipid droplets than fast-twitch fibres (Choi et al., 2016; Gueugneau et al., 2015). Therefore, since elderly people exhibit greater fat infiltration in both the multifidus and erector spinae muscles compared to younger individuals (Han et al. 2024; Liu et al. 2024), this factor may also partly explain the reduced endurance time of the back extensors observed for older adults.

Notably, the three studies reporting the greatest reductions in endurance time among elderly individuals (**Fig. 2**) employed different isometric fatiguing protocols: the standard Biering-Sørensen test (Mbada et al. 2011), a modified Sørensen test on a 45° Roman Chair (Bašič et al. 2013; Koch et al. 2025), and the Ito test (Han et al. 2024). Although these protocols impose varying mechanical demands on the back extensor muscles (Demoulin et al. 2006), the reduction in endurance time observed in older participants was consistently evident across tests. This finding suggests that the observed age-related decline in back extensor endurance is not attributable to the specific characteristics of a single test, but rather reflects a broader, more generalisable physiological decline associated with ageing. Nevertheless, with the exception of Parrella et al. (2025), none of the studies included in the meta-analysis quantified the force output during the endurance task, which was generally performed at an absolute intensity rather than relative to individual capacity. This methodological aspect may have influenced the magnitude of the age-related difference in endurance time, since tasks performed at an absolute force level may impose a higher relative workload on older adults due to their reduced maximal strength. However, even when controlling for force output—as in Parrella et al. (2025), who employed a task performed at 30% of participants' maximal voluntary isometric force—the forest plot (**Fig. 2**) still indicates a trend toward reduced endurance time with ageing. Additionally, studies that assessed general fatigue indicators, such as post-task force decline (Champagne et al. 2009; Parrella et al. 2025) and perceived exertion using the Borg scale (Champagne et al. 2009), reported no significant age-related differences in these parameters, suggesting that overall fatigue development was comparable between older and younger individuals. This is further supported by da Silva et al. (2015) who observed no group differences in Borg scores during two distinct submaximal fatiguing tasks not performed to exhaustion. Although only a few studies assessed these parameters, the findings suggest that the reduced endurance time observed in older adults might not be related to a greater general sense of fatigue, but rather to specific neuromuscular changes associated with ageing.

Potential neuromuscular mechanisms

From a neuromuscular perspective, five studies examined EMG parameters during fatiguing tasks for the back extensor muscles to better understand the mechanisms underlying age-related fatigue (Singh et al. 2011; Tsuboi et al. 2013; Kienbacher et al. 2014; da Silva et al. 2015; Parrella et al. 2025). A smaller decline in MF was reported in older adults during a 30-s fatiguing task at 80% MVC (Kienbacher et al. 2014). Similarly, a reduced decline in both MF and mean MPF was observed in older men during the Biering-Sørensen test until voluntary exhaustion (Tsuboi et al. 2013). These attenuated spectral shifts have been interpreted as potentially reflecting a greater proportion of type I muscle fibres with ageing. However, Tsuboi et al. (2013) did not observe a corresponding increase in back muscle endurance, unlike findings reported in other muscle groups (Avin and Law 2011). This discrepancy suggests that factors beyond muscle fibre-type composition may primarily underlie the age-related changes in EMG spectral shift, such as increased reliance on synergistic muscles. Older adults may recruit additional synergistic muscles more extensively than younger individuals to help sustain the endurance task, distributing the workload more broadly and consequently producing smaller EMG spectral shifts in the primary back extensor muscles. Moreover, Parrella et al. (2025) did not find significant differences in MFCV between age groups during a modified Biering-Sørensen test, further suggesting that intrinsic muscle fibre properties may not fully explain the attenuated spectral shifts observed in older adults. Taken together, these findings point toward alternative explanations, such as greater recruitment of synergistic muscles or altered load-sharing strategies in older individuals to maintain task performance. In this regard, older adults exhibited a different distribution of erector spinae muscle activity compared to young controls during fatigue. Specifically, they showed a more cranial activation pattern, which may reflect a compensatory strategy to redistribute load away from the lower lumbar segments—particularly the L4-L5 level—which are more susceptible to age-related degenerative changes (Hicks et al. 2009). In contrast, no differences in MF decline were reported during either a 60-s 45°

Roman chair exercise or a functional task involving holding a box in an upright position for 60 s (da Silva et al. 2015), possibly due to the lower intensity and shorter duration of these tasks compared to more demanding protocols employed in the other studies. Lastly, Singh et al. (2011) reported greater power in the 101-200 Hz frequency band of EMG signals in older adults compared to younger individuals during a 120-s isometric lumbar extension contraction at 60% MVC. However, no time-related changes were observed in any frequency band for either age group, leading the authors to suggest that this type of task likely did not induce significant fatigue-related changes in neuromuscular activation patterns. Overall, the variability in EMG outcomes and methodological approaches across studies limits the ability to draw definitive conclusions about the neuromuscular mechanisms underlying age-related fatigue of the back extensor muscles. In addition, it cannot be excluded that variability in EMG findings reflects not only methodological differences but also differences in motor strategies adopted by older adults. For example, to achieve the required task, older individuals may rely on different patterns of synergistic muscle recruitment, potentially influenced by task intensity, loading conditions, or individual compensatory mechanisms. Furthermore, to the best of the authors' knowledge, no animal studies have specifically assessed age-related changes in endurance or fatigue of these muscles, and therefore mechanistic insights from animal models cannot currently be applied to human back extensor fatigue.

Methodological considerations

Some important considerations should be made when interpreting age-related differences in muscle fatigue, which may guide future research in this area. First, future studies should extend beyond iso metric protocols to include dynamic tasks involving the spinal extensors, as age-related fatigue patterns may differ depending on the type of contraction (Avin and Law 2011). For instance, the assessment of isokinetic con tractions of the spinal extensor muscles may provide a more comprehensive understanding of how endurance capacity of these muscles changes with ageing. This is particularly relevant as such con tractions more closely reflect the movements per formed during activities of daily living compared with isometric contractions (Paris et al. 2022). Notably, the only study focusing on dynamic contractions of the back extensor muscles reported greater endurance in elderly individuals compared to young participants (Parreira et al. 2013), contrasting with the reduced endurance observed during isometric tasks in this review. However, future studies are needed to confirm or refute this finding. Additionally, physical activity levels should be assessed, as they may influence not only muscular endurance capacity but also an individual's predisposition to sustain prolonged effort. Among the included studies, only Han et al. (2024) and Kienbacher et al. (2014) quantified physical activity levels in their respective age groups. Lastly, sex differences in spinal extensor muscle fatigue should also be investigated, as previous research suggests that sex can be a factor influencing fatigue resistance during ageing (Allman and Rice 2002), but it remains unclear whether similar patterns extend to spinal muscles. Reporting both physical activity levels and sex differences is important as these variables could serve as potential moderating factors in future meta-analyses.

In conclusion, this systematic review and meta-analysis revealed that endurance of the back extensor muscles is reduced in older adults when performing isometric tasks. However, the specific neuromuscular mechanisms underlying this decline remain unclear due to mixed EMG findings and methodological variability across studies. Future studies should aim to clarify these mechanisms, potentially by employing protocols based on relative intensity (e.g., %MVC) rather than absolute loads and should also extend to dynamic contractions. Moreover, studies specifically examining age-related fatigue in the neck extensor muscles are warranted, as no such data were identified in the present review.

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CHAPTER 4

“Age-related alterations in trunk extensor force control during isometric and isokinetic contractions”

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4.1 Abstract

Background and objective: Although trunk extensor muscles are essential for spinal stability, research on age-related changes in force control of these muscles is limited, and the underlying neuromuscular mechanisms remain largely unexplored. This study aims to quantify the relationship between oscillations in lumbar erector spinae (LES) activity and torque fluctuations in young and older adults during isometric and isokinetic trunk extension contractions.

Methods: High-density surface electromyography (HDsEMG) signals were recorded from the right and left LES muscles of 20 young and 20 older participants using 13x5 electrode grids. Participants performed isometric and concentric trunk extensions on an isokinetic dynamometer at 25% and 50% of maximal voluntary contraction (MVC). Torque steadiness was quantified using the coefficient of variation (CoV) of torque. Coherence analysis in the δ band (0-5 Hz) was applied between the filtered interference HDsEMG and torque signals. Regional differences in HDsEMG-torque coherence were also assessed along the x- and y-axes of topographical coherence maps.

Results: Older individuals exhibited greater torque CoV than young adults during both isometric (+23.03%) and isokinetic (+72.62%) contractions ($p<0.001$), with the difference in isokinetic contractions being larger at 25% MVC (Group × Torque interaction; $p=0.007$). At this intensity, the older group also showed reduced HDsEMG-torque coherence (Group × Torque interaction; $p=0.004$). In contrast, coherence magnitude during isometric contractions was similar between groups ($p>0.05$), but older adults exhibited higher coherence in more cranial and medial LES regions ($p=0.005$ and $p=0.001$, respectively).

Conclusions: The results showed that older individuals exhibited the greatest impairment in force steadiness during low-intensity isokinetic contractions. Distinct neuromuscular patterns possibly influencing force stability emerged depending on contraction type.

4.2 Introduction

The force a muscle produces during voluntary movements depends on the number of motor units recruited and their discharge rates (Enoka and Duchateau 2017). However, even when a person attempts to apply a constant force, the output is not perfectly steady but fluctuates around an average value (Enoka and Farina 2021). These fluctuations are commonly quantified using magnitude-based measures, such as the standard deviation and coefficient of variation, which provide an index of force or torque steadiness (Enoka and Farina 2021).

Several studies have reported increased force fluctuations in older individuals during both isometric (Galganski et al. 1993; Tracy and Enoka 2002; Vanden Noven et al. 2014) and concentric or eccentric contractions (Burnett et al. 2000; Laidlaw et al. 2000; Keogh et al. 2006) of the upper and lower limb muscles. From a functional perspective, this suboptimal control of muscle force contributes to declines in physical performance, affecting a wide range of balance and mobility tasks (Kouzaki and Shinohara 2010; Chung-Hoon et al. 2016; Davis et al. 2020). Given these functional consequences, the neural mechanisms underlying force fluctuations have been a longstanding focus of research (Pethick et al. 2022).

It has been demonstrated that the low-frequency components (<10 Hz) of the neural drive to the muscle (i.e., the effective neural drive) are mainly responsible for force generation, as they reflect the common synaptic input to the motoneuron pool (Negro et al. 2009; Farina et al. 2014; Borzuola et al. 2025). This common input determines the neural command required for optimal force generation, while oscillations within this input challenge the stability of the force output (Farina and Negro 2015). Consequently, fluctuations in the effective neural drive determine force variability (Farina and Negro 2015). Accordingly, Castronovo et al. (2018), using high-density electromyography (HDsEMG) decomposition analysis, reported that reduced force steadiness in older individuals during submaximal contractions of the dorsiflexor muscles was associated with increased fluctuations in the common synaptic input to motoneurons.

However, the decomposition of HDsEMG signals remains challenging for some muscles, including the lumbar erector spinae. As an alternative, examining surface EMG (sEMG)/force relationship provides a useful approach to better understand the interaction between muscle activity and force production. Using this approach, previous studies have demonstrated that the low-frequency force fluctuations are correlated with the low-frequency components of the rectified interference sEMG (Yoshitake and Shinohara 2013; Moon et al. 2014). In addition, combining HDsEMG, which offers higher spatial sampling resolution than traditional sEMG, with principal component analysis (PCA), has been shown to enhance sEMG-based force estimation (Yoshitake and Shinohara 2013; Moon et al. 2014). PCA is a dimensionality-reduction technique that captures the common variability across multiple HDsEMG signals, selecting the most relevant components to generate a unique signal that explains most of the variance in the exerted force (Staudenmann et al. 2006). In addition, since HDsEMG allows the generation of topographical maps of muscle activity, oscillations in HDsEMG and torque signals can also be assessed in the spatial domain using maps that quantify the relationship between these two signals across the regions covered by the electrode grid. This approach helps determine whether specific muscle regions contribute more to the exerted torque than others (Arvanitidis et al. 2022). It has previously been used to assess force control of the trunk extensor muscles during both isometric and dynamic contractions in individuals with chronic low back pain (Arvanitidis et al. 2022, 2024b, a). However, this methodology has never been applied in older adults, where it could provide valuable insights into the effects of ageing on trunk extensor function.

The trunk is fundamental for performing activities of daily living, as it provides proximal stability for distal mobility (Forestieri Faccio et al. 2021). In particular, the trunk extensor muscles are the primary supportive muscles of the spine, counteracting the anterior flexion forces imposed by gravity, which is essential for stabilising the spine and maintaining an upright posture (Banno et al. 2019; Nakahira et al. 2025). Previous research on age-related differences in trunk extensor force steadiness has shown that older adults exhibit greater force fluctuations than younger individuals during a fatiguing task at 30% of maximal voluntary isometric force (Parrella et al. 2025) and a 15-s steadiness task at 10% of peak torque (Porto et al. 2020). In addition, older individuals with hyperkyphosis exhibited altered position and force sense (i.e., the ability to accurately reproduce a specific target position and force) in the trunk extensors, suggesting impaired proprioception (Keshavarzi et al. 2022). Notably, Forestieri Faccio et al. (2021) found that torque steadiness was overall greater in the trunk flexor muscles than in the trunk extensors of older individuals, suggesting that the force control of the trunk extensor muscles may be more affected by age. Moreover, Cangussu Oliveira et al. (2020) reported that torque steadiness of the trunk extensors showed the strongest association with vertebral fracture occurrence among various trunk muscle parameters, such as peak torque, rate of torque development and endurance.

Despite these findings, studies investigating force/torque steadiness of the trunk extensor muscles in ageing remain limited. Existing research has focused exclusively on isometric contractions, whereas dynamic contractions better resemble movements of daily activities (Paris et al. 2022). Moreover, the neural mechanisms underlying age-related alterations in force control in these muscles have yet to be elucidated.

This study aims to quantify the relationship between oscillations in lumbar erector spinae (LES) HDsEMG activity and torque fluctuations in the frequency domain in young and older adults during isometric and concentric trunk extension contractions. Additionally, regional differences in HDsEMG-torque coherence of the LES muscle will be analysed during both types of contractions. Considering the age-related alterations in motor unit properties and in the neural input to motor neurons (Pethick et al. 2022), it was hypothesised that older individuals would exhibit greater torque fluctuations during both isometric and dynamic contractions. Furthermore, we hypothesised that older individuals would have weaker correlation between HDsEMG and torque, accompanied by an altered regional distribution of LES HDsEMG-torque coherence.

4.3 Methods

4.3.1 Design and setting

This study employed a cross-sectional design and was approved by the Ethics Committee of the University of Birmingham (ERN_3410). All experimental procedures were conducted in accordance with the Declaration of Helsinki, and the study was reported following the STROBE guidelines (von Elm et al. 2008). Data collection took place between March and June 2025 at the Centre of Precision Rehabilitation for Spinal Pain (CPR Spine), University of Birmingham, UK. Participants attended a single experimental session lasting approximately 1.5 hours, and they provided a written informed consent prior to participation. They were encouraged to raise any questions or concerns regarding the study before providing consent.

4.3.2 Participants

Twenty older volunteers and twenty younger volunteers were enrolled in the study. The sample size was determined a priori based on a statistical power analysis (G*Power statistical software v.3.1.9.4) for a mixed-model ANOVA (within-between interaction) with α=0.05, power (1–β err prob)=0.80, and effect size=0.25. These parameters were based on previous similar studies investigating torque steadiness and HDsEMG-torque coherence, which reported small to moderate effect sizes (Arvanitidis et al. 2022; Parrella et al. 2025). Young participants (age range: 18-35 years) were included if they did not have a history of neurological or orthopaedic disorders, while older participants (age range: 65-80 years) were included if they met the criteria to be defined as “medically stable” for exercise studies (Greig et al. 1994). In addition, both young and older volunteers were excluded from the study if they had a history of chronic low back pain, current low back pain, lumbar radiculopathy, spinal surgery or spinal deformities (i.e., scoliosis, spondylolisthesis, spondylolysis) (Parrella et al. 2025). Participants were recruited from the local Birmingham area and the University of Birmingham student and staff communities via social media advertisements, posted information leaflets, and word of mouth. In addition, older participants were recruited through the Birmingham 1000 Elders program, a registry of healthy older adults maintained by the university for participation in research studies.

4.3.3 Functional assessment

Before starting the session, blood pressure was measured in older participants to ensure they were fit to participate. Participants were excluded from testing if their resting diastolic blood pressure was ≥100 mmHg and/or their resting systolic blood pressure was ≥160 mmHg (Giffin et al. 2020). If blood pressure values were within the acceptable range, a functional assessment was performed using the Berg Balance Scale (BBS). The BBS consists of fourteen functional tasks of increasing difficulty, ranging from quiet stance, sit-to-stand, weight shifting, and reaching, to turning in place, tandem stance, and single-leg stance (Berg et al. 1992; Downs et al. 2014). Each task was scored on a 5-point scale from 0 (unable to perform) to 4 (performed independently) according to standardised criteria. The maximum total score was 56 points, with higher scores indicating better balance and the absence of detectable balance difficulties. Participants were allowed to rest as needed between tasks.

4.3.4 Questionnaires

Participants of both groups were asked to complete a questionnaire to assess their physical activity (PA) level. Specifically, young adults completed the Baecke Physical Activity Questionnaire (BPAQ), a validated self-report instrument consisting of sixteen questions that assess PA during a typical week over the past year across three domains: (1) work (eight questions), (2) sport/exercise (four questions), and (3) leisure-time non-sport activity (four questions) (Baecke et al. 1982; Healey et al. 2020). In the sport/exercise domain, participants answered three additional questions regarding the type of activity, the average number of hours per week, and the number of months per year they engaged in that activity. Except for questions about main occupation and type of sport/exercise that had three response options (i.e., low, moderate, or high intensity/activity), all other items were rated on a five-point Likert scale, ranging from never to always/very often. Each domain yielded a score between 1 and 5, which were then summed to produce a total physical activity score ranging from 3 (minimum) to 15 (maximum).

Since the original version of the BPAQ has been shown to be less reliable in elderly populations (Tebar et al. 2022), an adapted version, the Modified BPAQ, was used for the older group (Voorrips et al. 1991). This version assesses three domains: (1) household activities, (2) sport/exercise, and (3) leisure-time activities. The household domain consists of ten questions, each with four to five response options. For the sport/exercise and leisure-time domains, participants reported the type of activity (up to two sports and up to six leisure-time activities), specifying the intensity, hours per week, and months per year for each activity. Scores from the household, sport/exercise, and leisure-time domains were summed to produce a continuous, unitless total physical activity score.

The total scores from both the BPAQ and the Modified BPAQ were divided into tertiles based on group-specific cutoffs to classify participants as having low, moderate, or high habitual physical activity, in line with previous studies (Sandercock et al. 2008; Hertogh et al. 2008; Tebar et al. 2022).

4.3.5 Dynamometer setup

The torque exerted by participants during maximal voluntary contractions (MVCs) and submaximal torque steadiness tasks was measured using an isokinetic dynamometer (System 3 Pro, Biodex

Medical Systems, New York). All contractions were performed on the Biodex Dual Position Back Extension/Flexion Attachment, with participants' knees flexed at 90° and their feet positioned parallel to the floor at a distance equal to the inter-acromial distance (**Figure 1**). The front of the seat was tilted approximately 15° clockwise, and the height of the chair was adjusted so that the rotational axis of the dynamometer was aligned bilaterally with the anterior superior iliac spines of participants. This position, referred to as the compressed isolated lumbar position, was selected to maximise the contribution of LES muscle to the resultant torque (Arvanitidis et al. 2022). Participants' upper trunk, thighs, and pelvis were also securely strapped to the seat to minimise compensatory movements during the contractions, and a specific attachment was used to reduce involvement of the knee muscles.

For the isometric trunk extension contractions, the dynamometer was locked at a hip-to-trunk angle of 90° (Arvanitidis et al. 2022) (**Figure 1a**). In contrast, concentric trunk extension contractions were performed in isokinetic mode over a total range of motion of 50°, starting at 30° of hip-to-trunk flexion (**Figure 1b**) and ending at 20° of hip-to-trunk extension (**Figure 1c**). This setup was chosen to isolate lumbar motion and minimise compensatory movements from the lower extremities (Arvanitidis et al. 2024b). The angular velocity for the extension phase was set at $10°\cdot s^{-1}$, while it was set at $60°\cdot s^{-1}$ when returning to the starting position. However, the return phase was performed passively by the researcher, who repositioned the chair after each repetition. The same setup was used for both MVCs and submaximal tasks for each type of contraction (isometric or isokinetic). For all participants, the offset due to trunk weight was consistently removed with the trunk positioned at a specific angle.

4.3.6 Testing protocol

Participants were first familiarised with the dynamometer to warm up the trunk extensor muscles. The testing protocol consisted of two parts: isometric contractions followed by isokinetic concentric contractions.

For the isometric part, the warm-up consisted of three sets of eight low-to-moderate intensity isometric contractions, with one minute of rest between sets. Participants then performed three isometric MVCs, separated by 2 minutes of rest. The highest maximal voluntary torque obtained during these three MVCs was used to determine the target torque for the submaximal tasks. After completing a few practice trials, participants performed two sustained isometric trunk extension contractions at 25% MVC (2.5-s ramp-up, 30-s hold, 2.5-s ramp-down) and two at 50% MVC (5-s ramp-up, 15-s hold, 5-s ramp-down). A rest period of 30 s was provided between contractions at 25% MVC and 1 min between contractions at 50% MVC. The order of torque levels was randomised, with a 2-min rest period separating the two intensities. After the submaximal tasks, the isometric MVC test was performed again to rule out the presence of muscle fatigue.

For the isokinetic part, the protocol followed essentially the same structure as the isometric one. The warm-up consisted of three sets of five low-to-moderate intensity contractions performed across the full 50° range of motion (from 30° hip-to-trunk flexion to 20° extension), with one minute of rest between sets. Participants then performed three isokinetic MVCs, separated by 2 minutes of rest, and the highest torque obtained was used for the submaximal tasks. After a few practice trials, participants performed four submaximal concentric trunk extension contractions at 25% MVC and four at 50% MVC. With an angular velocity of $10°\cdot s^{-1}$ over a 50° range of motion, each contraction lasted 5 s. During these tasks, unlike isometric ones, no ramped contractions were used. Instead, participants were instructed to maintain their exerted torque, displayed as a bar, aligned with a single target line (25% or 50% MVC) throughout the range of motion. Rest periods of 15 s and 30 s were provided between contractions at 25% and 50% MVC, respectively, with the order of intensities randomised and separated by a 2-min rest period. An additional isokinetic MVC test was performed after the submaximal tasks again to rule out the presence of muscle fatigue.

During all submaximal contractions (both isometric and isokinetic), participants received real-time visual feedback of the target torque on a computer monitor positioned 1.5 m in front of them (**Figure 1**). They were instructed to reach the specified % MVC target (25% or 50%) and maintain their torque as steadily as possible for the duration of the task. Instructions were provided prior to the contractions. Participants received verbal encouragement to exert maximal effort during the MVCs, whereas no feedback was given during the submaximal tasks, and the testing environment was kept quiet.

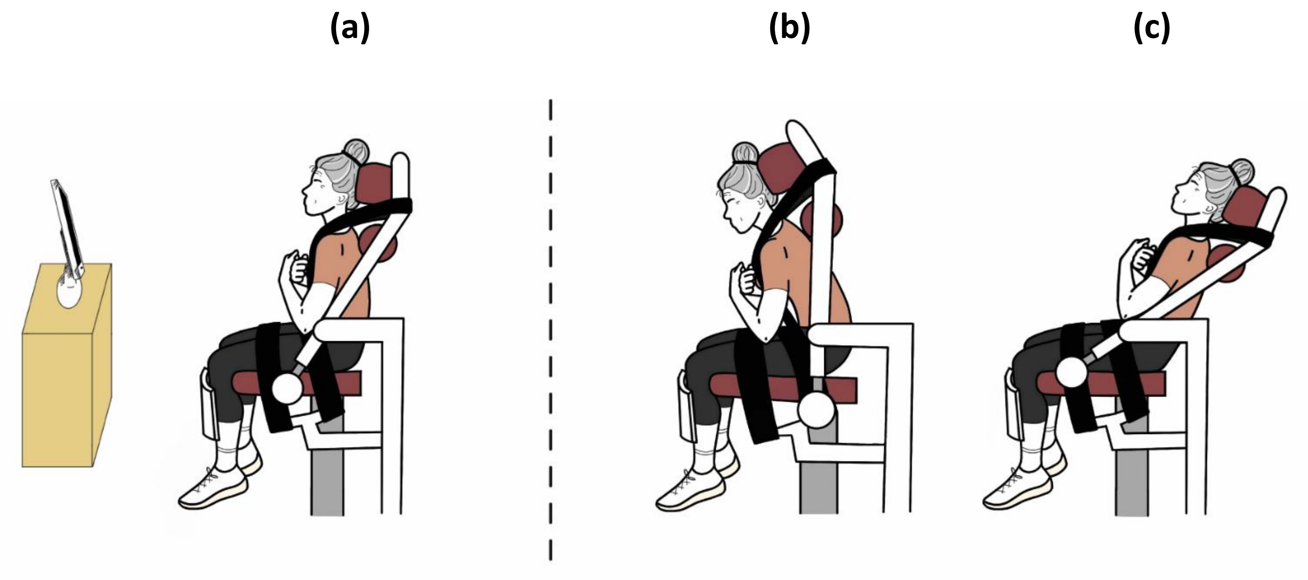


Figure 1. Representation of the experimental setup. The participant is seated on the Biodex chair during **(a)** isometric trunk extension tasks, with the hip-to-trunk angle set at 90°, and during isokinetic trunk extension tasks, starting from **(b)** 30° of hip-to-trunk flexion and ending at **(c)** 20° of hip-to-trunk extension.

4.3.7 HDsEMG signal recording

The CEDE checklist was used to ensure accurate reporting of the acquisition and processing of EMG data (Besomi et al. 2024). HDsEMG signals were recorded bilaterally in monopolar mode from the LES muscles during both MVCs and submaximal contractions, using adhesive grids of 64 equally spaced electrodes (GR08MM1305, gold-coated, 1 mm diameter, 8 mm interelectrode distance; OT Bioelettronica, Turin, Italy). Each grid consisted of 13 rows and 5 columns of electrodes with a missing electrode in the upper left corner. Before electrode placement, a double-sided adhesive foam (FOA08MM1305; OT Bioelettronica) was applied to the grids, and the electrode cavities were filled with conductive paste (AC-CREAM; SPES Medica, Genoa, Italy) to ensure proper electrode-skin contact. After identifying the perimeter of the right and left LES muscles by manual palpation, participant's skin was shaved, gently abraded using abrasive paste and cleaned with water to ensure the best conductivity of HDsEMG signals. According to Falla et al. (2014), the two grids were placed ~2 cm lateral to the L5 spinous process mid-point, covering the low back approximately from L5 to L2 on both the right and left sides (**Figure 2**). To reduce electrical noise and improve signal quality, three reference electrodes were placed on participants: one over a lumbar or thoracic vertebra and one on each wrist. The HDsEMG signals were synchronised with the torque signals from the isokinetic dynamometer via the auxiliary input of the HDsEMG amplifier (Quattrocento, OT Bioelettronica, Torino, Italy). All HDsEMG and torque signals were amplified (x150), sampled at 2048 Hz, band-pass filtered (10-500 Hz, first order, -3 dB), and digitised with a 16-bit A/D converter. Recordings were performed using the OTBiolab+ software.

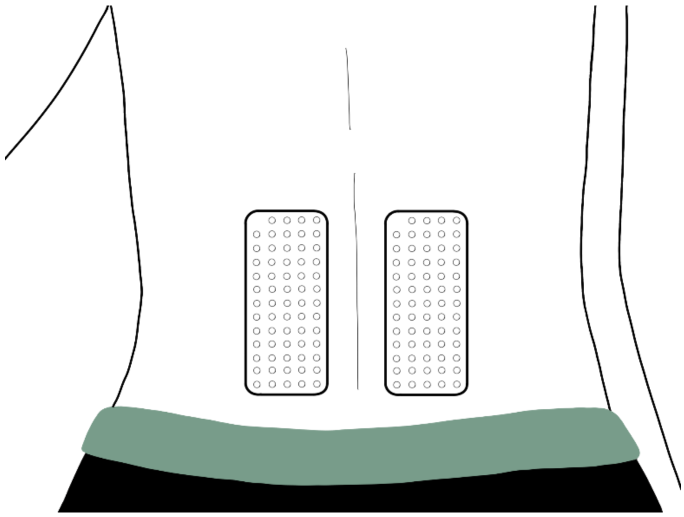


Figure 2. Grids of electrodes placed over the right and left lumbar erector spinae (LES) muscles of participants.

4.3.8 PCA

PCA was applied to reduce the high dimensionality of the HDsEMG data, which can introduce redundant or correlated information in the recorded signals, such as noise, artefacts, or channels with low muscle activity (Naik et al. 2016). In particular, PCA was implemented to reduce the large set of correlated sEMG signals into a smaller subset of uncorrelated variables defined principal components (PCs) (Arvanitidis et al. 2022). According to Naik et al. (2016), PCs capturing more than 80% of the data variance are typically retained. In the present study, this threshold was set at 85% of total variance.

From each of the two HDsEMG grids (i.e., on the right and left LES muscles), 59 adjacent bipolar channels were derived from the 64 monopolar signals, resulting in twelve longitudinal bipolar recordings per column, except for the far-left column, which contained eleven electrode pairs due to the missing electrode. Prior to coherence analysis, the bipolar HDsEMG channels underwent the following pre-processing steps for PCA-based dimensionality reduction (Staudenmann et al. 2006; Arvanitidis et al. 2022): (1) high-pass filtering at 10 Hz; (2) selection of the most informative subset of PCs obtained by applying PCA to the 59 differential HDsEMG signals in the time domain; (3) full-wave rectification and averaging of the selected PCs to generate a single time-domain signal for each muscle; (4) additional low-pass filtering at 10 Hz to capture slow-frequency fluctuations in motor unit firing rate/recruitment (Negro et al. 2009), (5) smoothing with a first-order Savitzky-Golay filter; and (6) removal of the DC (zero-frequency) components. These processing procedures resulted in the Final Signal Envelope, obtained by applying PCA to the HDsEMG grid, which contained the relevant force-related information (i.e., low-frequency components of the HDsEMG data).

After extracting the Final Signal Envelope, coherence analysis was performed to quantify the similarity between the HDsEMG envelope and the torque signals. This analysis was conducted for all submaximal contractions (both isometric and isokinetic), with similarity calculated for each repetition and then averaged to yield a single coherence value at a given torque level.

4.3.9 Coherence analysis

Coherence analysis was performed to indirectly estimate the strength of common rhythmic synaptic inputs across the motor unit pool and to examine their association with torque (Arvanitidis et al. 2022). The coherence between HDsEMG envelope and torque signals was computed using the magnitude-squared coherence (MSC) method with a 1-s Hamming window and 50% overlap, as previously reported (Castronovo et al. 2015). In particular, the Final Signal Envelope obtained from the PCA was correlated with the torque signal in the frequency domain. The MSC approach is frequently used to quantify the linear correlation between two signals in the frequency domain, ranging from 0 (no correlation) to 1 (perfect correlation). In this study, MSC was computed using the *mscohere* function of the MATLAB Signal Processing Toolbox.

Coherence analysis focused exclusively on the δ band (0-5 Hz), as this frequency range is most relevant for force generation (Farina and Negro 2015). Coherence within this band was quantified by integrating the coherence estimates across the δ frequency range. To enable statistical comparison, coherence estimates (C) were then converted to Fisher's z values (FZ), as indicated in the equation below (Castronovo et al. 2015). Since coherence can be influenced by crosstalk commonly contaminating sEMG recordings, the bias was empirically determined and removed. This bias was defined as the maximum value of the coherence profile at 250 Hz, a frequency range where no significant correlated activity is expected (Castronovo et al. 2015).

$$FZ = atanh\left(\sqrt{C}\right) - bias$$

Additionally, topographical maps of coherence were generated, as previously described (Arvanitidis et al. 2022, 2024b). In these maps, each of the 59 bipolar recordings within each HDsEMG grid was individually assessed for coherence with the filtered torque signal. This procedure yielded 59 coherence values in the δ band per grid, which were then used to construct the topographical coherence maps. These maps were then normalised to the maximum coherence observed at each torque level (25% or 50% MVC). The centroid of coherence was calculated to determine its spatial location along the medial-lateral (x-axis) and cranial-caudal (y-axis) directions. This approach allowed us to assess whether specific muscle regions contributed more to torque generation.

Coherence variables described above were extracted from the steady portion of the submaximal contractions using a non-overlapping 0.5 s sliding averaging window with a custom MATLAB script (MATLAB 2022b, Mathworks Inc., Natick, MA, USA). The script plotted the torque signal of each participant, and the steady portion of each contraction was identified to determine the start and end points for analysis. All variables were computed separately for each repetition and then averaged to obtain a single value for each torque level within each contraction condition (isometric or isokinetic).

4.3.10 RMS

For these calculations, the 59 adjacent bipolar channels from each grid were processed differently from the PCA procedure described above, in order to compute the root mean square (RMS) of the

HDsEMG signals. Prior to these analyses, HDsEMG signals were band-pass filtered (10-350 Hz, second-order, zero-lag Butterworth) (Martinez-Valdes et al. 2019). The signal quality of each channel was also visually inspected, and channels with a low signal-to-noise ratio were excluded. As the removal rate was below 15% in this study, RMS values could be estimated reliably from all submaximal trunk extension tasks, including both isometric and isokinetic contractions (Arvanitidis et al. 2021).

RMS values were computed for each bipolar channel of both HDsEMG grids. The 59 RMS values obtained from each grid were then averaged to yield a single representative value of the global myoelectric activity (RMSmean) for each muscle during each task. Specifically, as with the coherence variables, RMSmean values were extracted from the steady portion of the submaximal contractions, and were computed separately for each repetition. The values were then averaged to obtain a single value for each torque level within each contraction condition (isometric or isokinetic).

To allow statistical comparison between participants, the RMSmean values were normalised to the maximum RMS (RMSmax) obtained during the MVC tests, resulting in normalised RMS values (RMSnorm). In particular, RMSmean values from the submaximal isometric tasks were normalised to the RMSmax recorded during the highest isometric MVC, whereas RMSmean values from the submaximal isokinetic tasks were normalised to the RMSmax obtained during the highest isokinetic MVC. A 200-ms window around the peak torque was used to extract the RMSmax values during the MVC of each type of contraction.

4.3.11 Torque signal analysis

For each participant, the highest peak torque obtained during the isometric MVC and the highest peak torque obtained during the isokinetic MVC (SI unit: N·m) were used as measures of maximal trunk extension strength. To enable statistical comparisons between groups, both torque values were normalised to each participant's body mass (N·m·kg-1) (Jaric et al. 2002). During the isometric and isokinetic submaximal tasks, torque steadiness was quantified as the amplitude of torque fluctuations, expressed in relative terms by the coefficient of variation of the torque signal (CoV = SD/mean × 100) over time. The CoV of torque was calculated separately for each repetition, and the resulting values were averaged to obtain a single value for each torque level within each contraction condition (isometric or isokinetic). The same time window selected for the HDsEMG analysis was used for this calculation.

4.3.12 Statistical analysis

Statistical analysis was performed using IBM SPSS 24.0 (IBM Corp., Armonk, NY, United States) and Jamovi 1.6.23 (The Jamovi project, Sydney, Australia), while graphs were generated in R (R Development Core Team, 2023). The Shapiro-Wilk test was used to check normality of the data. When normality was not assumed, nonparametric tests were applied. Independent t-tests were conducted to compare young and older participants in body mass and BMI, while the Mann-Whitney U test was used to assess group differences in height.

For the isometric contractions, the Mann-Whitney U test was used to assess group differences in baseline isometric MVC values (MVCpre_ISO). The Wilcoxon signed-rank test was then applied to evaluate within-group changes between MVCpre_ISO and post-task values (MVCpost_ISO) following the isometric submaximal contractions. Linear Mixed Models (LMMs) (Jiang and Nguyen 2021) were applied to the RMSnorm values. In contrast, Generalised Linear Mixed Models (GLMMs) (Jiang and Nguyen 2021) were used for the HDsEMG-torque coherence values (z-coherence), the centroid of coherence on both axis (coher_x and coher_y), and torque variables (torque_mean and torque CoV), as the residuals of these variables did not meet the assumption of normality.

For the isokinetic contractions, an independent t-test was conducted to compare young and older participants in baseline isokinetic MVC values (MVCpre_DYN). Paired-samples t-tests were then performed to examine within-group differences between MVCpre_DYN and post-task values (MVCpost_DYN) following the isokinetic submaximal contractions. LMMs were applied to RMSnorm, coher_x, and coher_y, whereas GLMMs were used for z-coherence, torque_mean and torque CoV.

Group (young vs. older), Torque (25% vs. 50%), and Muscle (right LES vs. left LES) were included as fixed effects in all LMMs and GLMMs for both contraction types, with participant ID entered as a random intercept to account for within-subject variability. However, muscle side was not considered for torque-related variables, as torque values were obtained from the global output of both muscles. The LMMs were fitted using restricted maximum likelihood estimation (REML), and the significance of fixed effects was assessed using Satterthwaite's approximation for the degrees of freedom. For the GLMM, a gamma distribution with a log link function was used for right-skewed data, whereas a gamma distribution with an inverse link function was used for left-skewed data.

For all analyses, the significance level (α) was set to 0.05. When multiple comparisons were performed, p-values were adjusted using the Holm-Bonferroni correction. The data are reported as mean ± Standard Deviation (SD) for parametric data and as median and interquartile range (IQR) for nonparametric data. All values reported in the Results represent raw descriptive data, while adjusted (model-based) means were not reported.

4.4 Results

4.4.1 Participants' characteristics

No significant differences were found in body mass, height and BMI between young and older participants ($p > 0.05$). Descriptive characteristics of all participants are summarised in **Table 1**.

Instead, PA levels are summarised in **Table 2**, which presents the score ranges for each tertile (low, moderate and high PA) separately for each group, along with the percentage of participants in each tertile.

Characteristic	**Elderly** (*n* = 20; 12 F, 8 M)	**Young** (*n* = 20; 12 F, 8 M)	***p* value**
Age (years)	74.80 ± 4.74	24.75 ± 5.00	-
Body mass (kg)	65.05 ± 11.56	64.15 ± 11.08	*0.803*
Height (m)	1.65 (0.24)	1.69 (0.14)	*0.314*
BMI (kg/m^2)	23.06 ± 2.54	21.90 ± 2.69	*0.169*
BBS score	54.75 ± 1.33	/	-
MVC_{pre}_ISO ($N \cdot m \cdot kg^{-1}$)	2.08 (1.56)	2.89 (1.46)	*0.013**
MVC_{pre}_DYN ($N \cdot m \cdot kg^{-1}$)	2.24 ± 0.74	2.99 ± 0.74	*0.003**

Table 1. Characteristics of all participants separated by group. *BMI* body mass index. *PA level* physical activity level. *BBS* Berg Balance Scale. *MVCpre_ISO* baseline isometric maximal voluntary contraction. *MVCpre_DYN* baseline isokinetic maximal voluntary contraction. Data are presented as mean ± SD for all variables except for height and MVCpre_ISO (median and IQR). The asterisk (*) indicates statistically significant differences between groups.

	Low PA level (tertile 1)	**Moderate PA level (tertile 2)**	**High PA level (tertile 3)**
Elderly	≤ 10.05 *(35%)*	10.06 - 18.74 *(30%)*	> 18.74 *(35%)*
Young	≤ 7.13 *(40%)*	7.14 - 8.68 *(25%)*	> 8.68 *(35%)*

Table 2. Physical activity (PA) levels in each group, showing score ranges for tertiles defining low, moderate, and high PA, with participant percentages in brackets.

4.4.2 ISOMETRIC CONTRACTIONS

4.4.2a MVC

MVC_{pre}_ISO values were significantly lower in older compared to younger participants ($p < 0.05$) (**Table 1**). In addition, within each group, no significant differences were observed between

MVC$_{pre}$_ISO and MVC$_{post}$_ISO (young: p = 1.000; elderly: p = 0.073), indicating no insurgence of muscle fatigue.

4.4.2b Torque steadiness

A significant main effect of Group was found for torque CoV ($X^2 = 21.164$; $p < 0.001$), indicating overall higher CoV values in older individuals (**Figure 3a**). Specifically, older adults exhibited 23.03% greater CoV relative to younger individuals. Lastly, the exerted torque_mean at the two torque levels was similar between groups ($X^2 = 0.318$; $p = 0.573$). No significant interactions were found for either torque CoV or torque_mean ($p > 0.05$).

4.4.2c RMSnorm

The fixed-effects omnibus test revealed a significant main effect of Group ($F = 92.956$; $p < 0.001$), with older participants exhibiting overall greater RMSnorm values than younger participants (**Figure 3b**). A significant main effect of Torque was also found ($F = 37.077$; $p < 0.001$), with RMSnorm values being higher at 50% MVC compared to 25% MVC. No significant interactions were found ($p > 0.05$).

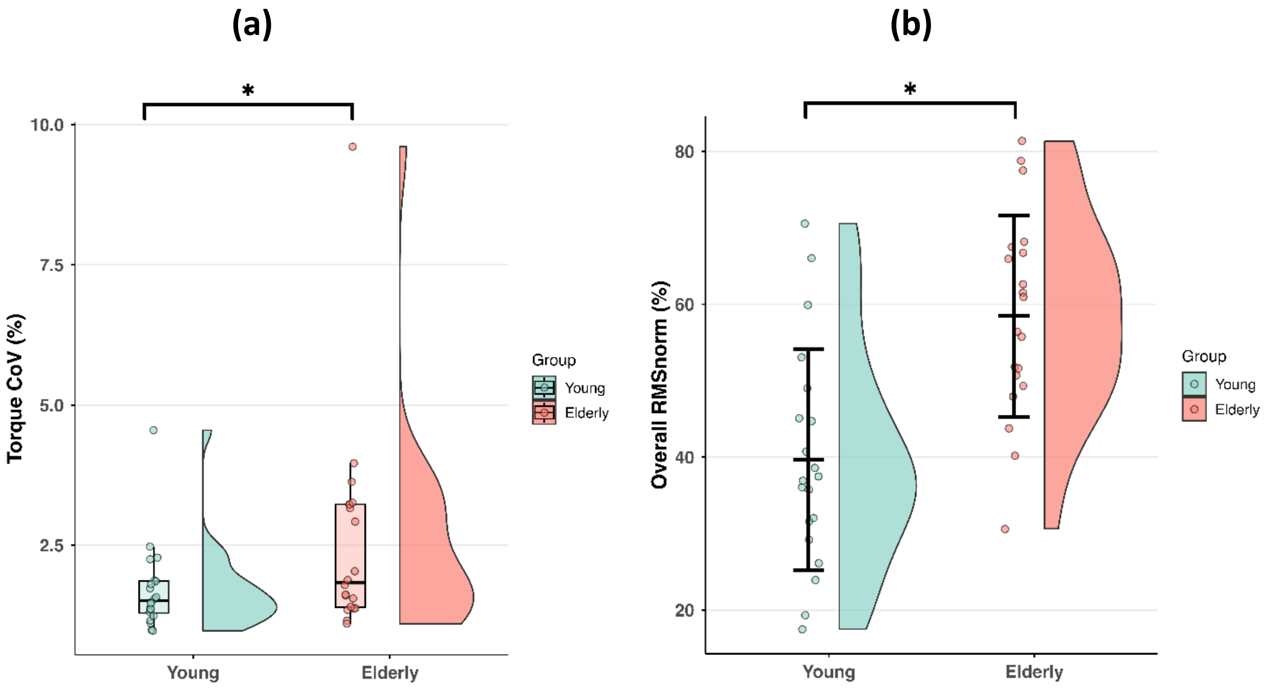
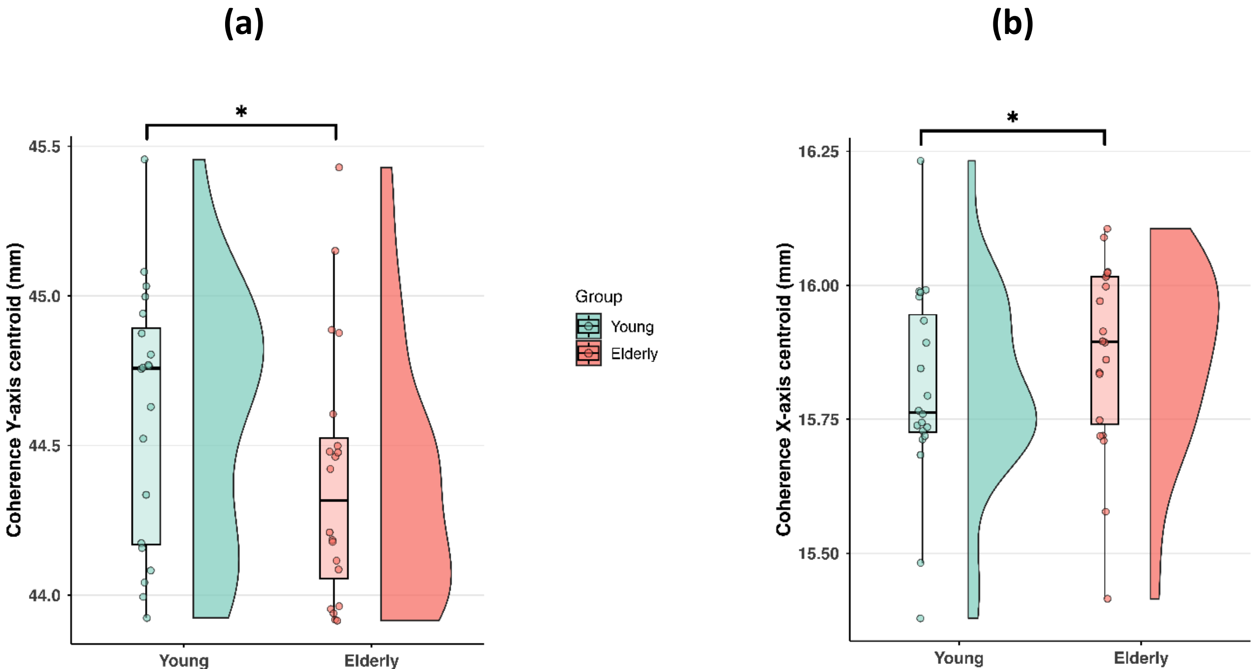


Figure 3. Coefficient of Variation (CoV) of torque **(a)**, overall normalised root mean square (overall RMSnorm) values **(b)** in young and elderly participants during the isometric trunk extension contractions. Median (IQR) for torque CoV. Mean ± SD for overall RMSnorm values. The distributions are illustrated using half-violin plots with embedded boxplots and individual data points for torque CoV, and mean ± SD plots with individual data points for overall RMSnorm values; * Main effect of Group.

4.4.2d Coherence

A significant main effect of Group on the centroid of HDsEMG-torque coherence along the y-axis was observed (X^2 = 7.978; p = 0.005), indicating that HDsEMG-torque coherence was higher in more cranial regions of the LES muscles of older participants compared to younger individuals (**Figure 4a**). A significant main effect of Group was also found for the centroid of coherence along the x-axis (X^2 = 10.343; p = 0.001), with older participants showing higher HDsEMG-torque coherence in more medial regions of the LES muscles compared to younger individuals (**Figure 4b**). No significant interactions were found for the centroid of coherence along either the x- or y-axis (p > 0.05). Topographical maps showing HDsEMG-torque coherence distribution for both groups are presented in **Figure 4c**. Lastly, no significant main effects or interactions were found for z-coherence values (young: 1.62 (0.27); elderly: 1.60 (0.25); p > 0.05).



(c)

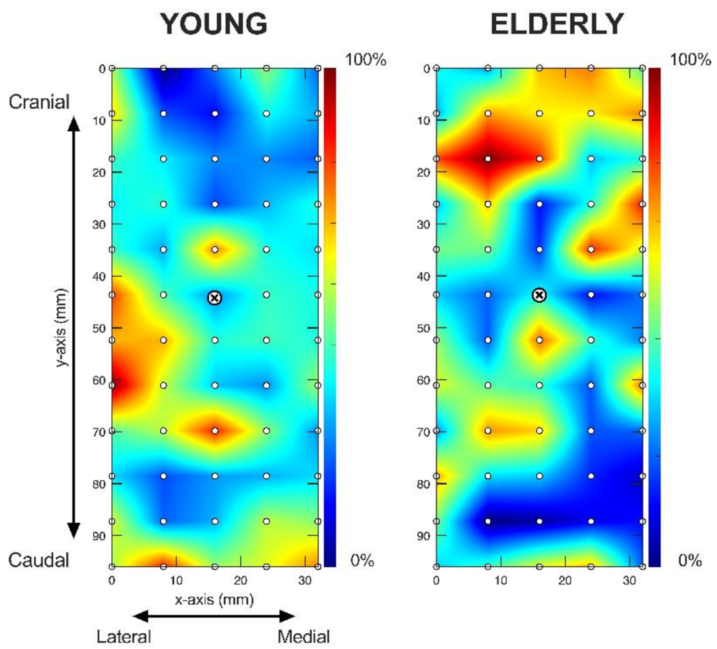


Figure 4. Centroid of high-density surface EMG (HDsEMG)-torque coherence along the y-axis **(a)** and along the x-axis **(b)** in young and elderly participants during the isometric trunk extension contractions. Median (IQR). The distributions are illustrated using half-violin plots with embedded boxplots and individual data points; ***** Main effect of Group. Representative topographical maps of coherence for one young and one older participant **(c)**. The white circle in the middle of the maps represents the centroid of coherence. The spine is located towards the medial side of the x-axis.

4.4.3 ISOKINETIC CONTRACTIONS

4.4.3a MVC

MVCpre_DYN values were significantly lower in older compared to younger participants ($p < 0.05$) (**Table 1**). In addition, within each group, no significant differences were observed between MVCpre_DYN and MVCpost_DYN (young: $p = 0.131$; elderly: $p = 0.490$), indicating no insurgence of muscle fatigue.

4.4.3b Torque steadiness

A significant main effect of Group was found for torque CoV (X^2 = 89.470; $p < 0.001$), indicating overall higher values in older individuals (**Figure 5a**). Specifically, older adults exhibited 72.62% greater CoV relative to younger individuals. A significant Group x Torque interaction was also found (X^2 = 7.290; $p = 0.007$) (**Figure 5b**). Post-hoc analyses revealed significantly greater CoV values in the older group compared to the younger group at both 25% and 50% MVC ($p < 0.001$). However, torque CoV was 107.84% higher at 25% MVC and 61.37% higher at 50% MVC in older individuals compared to the younger group, indicating that group differences were more pronounced at lower contraction intensities. Lastly, the exerted torque_mean at the two torque levels was similar between groups (X^2 = 0.412; $p = 0.521$).

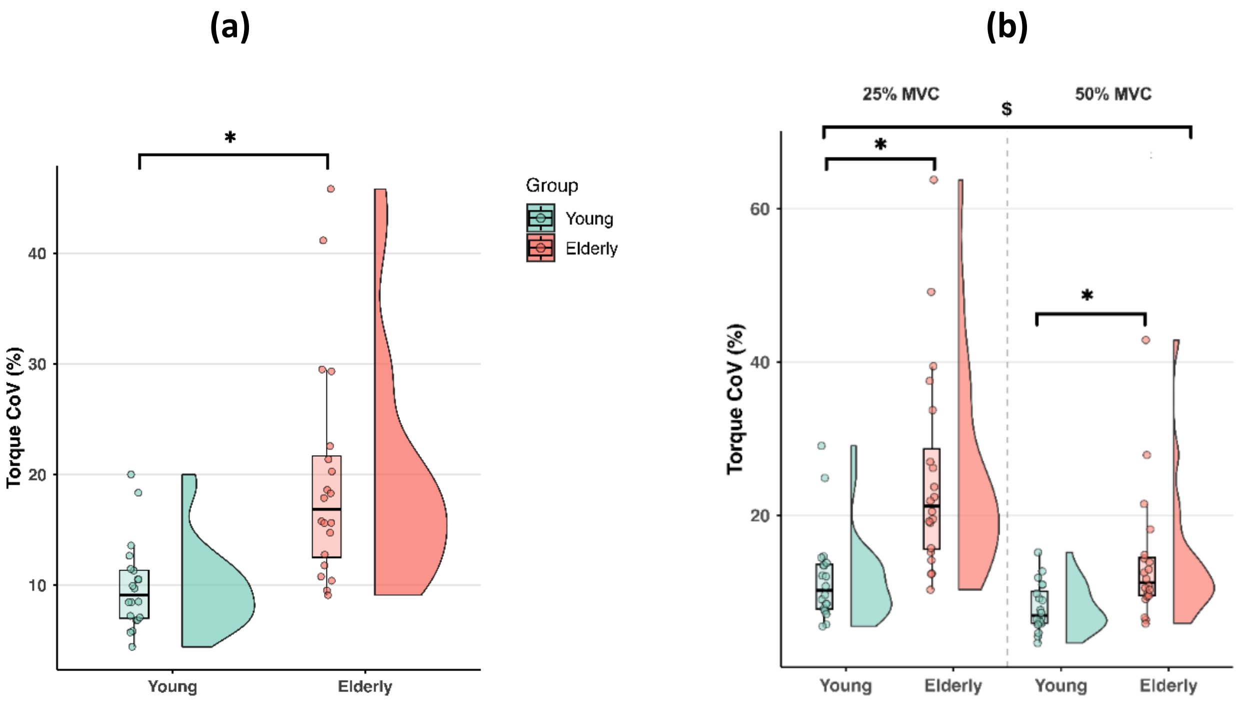
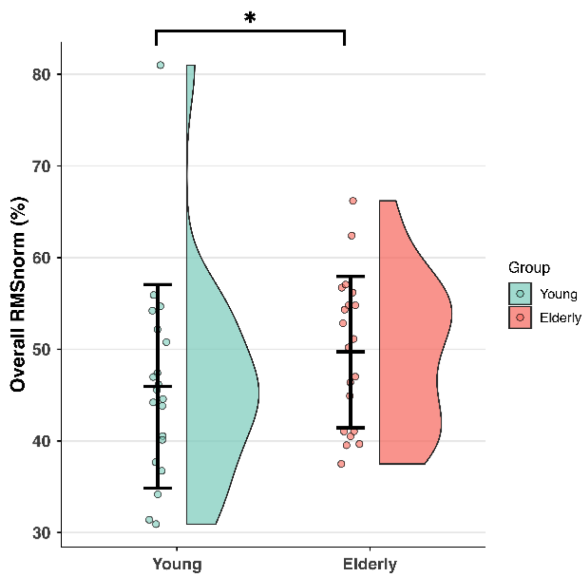


Figure 5. Overall Coefficient of Variation (CoV) of torque **(a)**, and torque CoV at 25% and 50% of maximal voluntary contraction (MVC) **(b)** in young and elderly participants during the isokinetic trunk extension contractions. Median (IQR). The distributions are illustrated using half-violin plots with embedded boxplots and individual data points; ***** Main effect of Group. **$** Group x Torque interaction.

4.4.3c RMSnorm

A significant main effect of Group was observed for RMSnorm values ($F = 8.151$; $p = 0.005$), with older participants exhibiting overall higher RMSnorm values compared to younger participants (**Figure 6a**). RMSnorm values also showed a significant main effect of Torque ($F = 102.091$; $p < 0.001$), being greater at 50% MVC than at 25% MVC. In addition, a significant Group x Muscle interaction was found ($F = 5.163$; $p = 0.025$) (**Figure 6b**). Post-hoc comparisons revealed that older participants showed significantly greater RMSnorm values in the right LES compared to the young group ($p = 0.002$), but not in the left LES ($p > 0.05$). A trend toward higher values in the right than left LES within the older group was also found ($p = 0.065$).

(a)



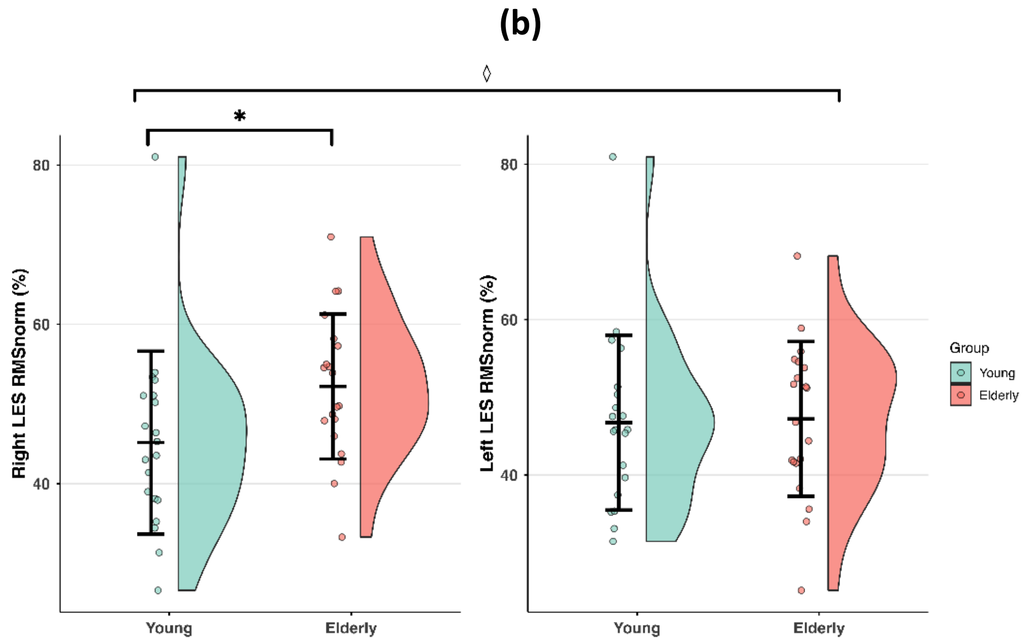
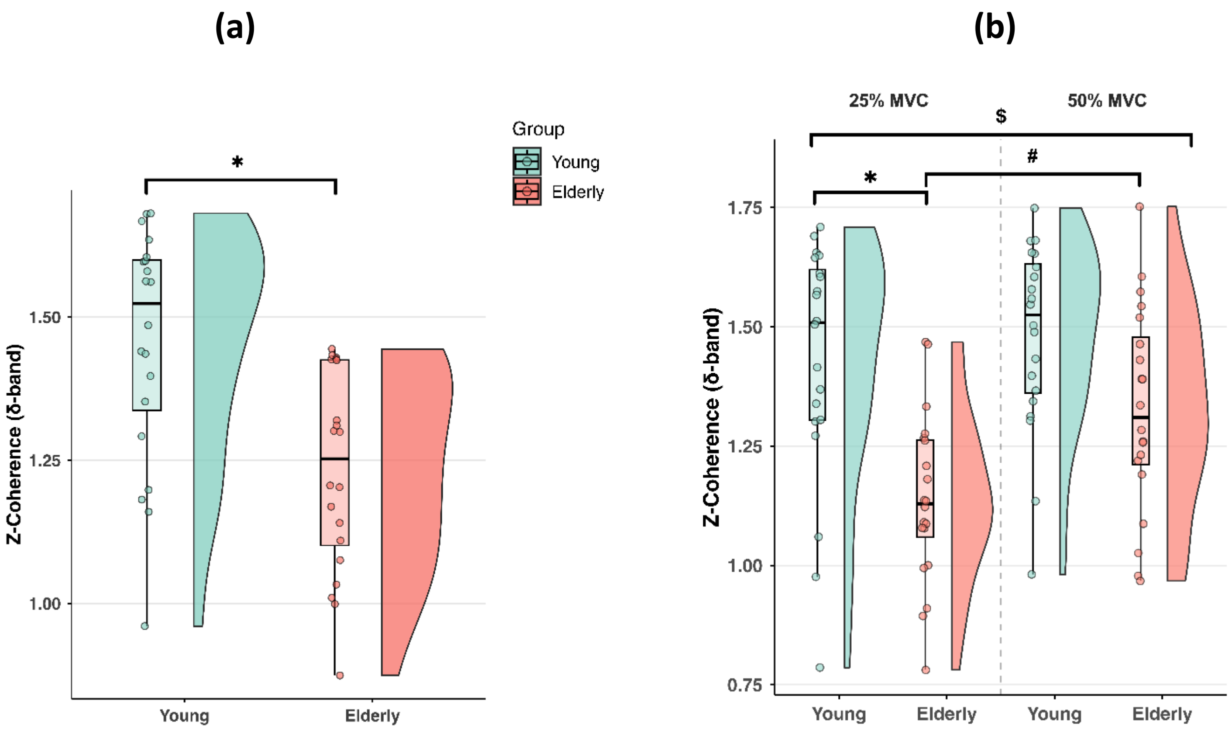


Figure 6. Overall normalised root mean square (RMSnorm) values **(a)**, and RMSnorm values in the right and left lumbar erector spinae (LES) **(b)** in young and elderly participants during the isokinetic trunk extension contractions. Mean ± SD. The distributions are illustrated using half-violin plots with embedded mean ± SD plots with individual data points; ***** Main effect of Group. ◊ Group x Muscle interaction.

4.4.3d Coherence

No significant main effects or interactions were found for the centroid of coherence along the y-axis ($p > 0.05$). In contrast, a significant main effect of Muscle was observed for the centroid of coherence along the x-axis ($F = 85.681$; $p < 0.001$), with the left LES exhibiting coherence that was overall more medial compared to the right LES. A significant main effect of Group effect was observed for z-coherence values ($X^2 = 25.819$; $p < 0.001$), with older participants showing overall lower coherence values compared to younger individuals (**Figure 7a**). A main effect of Torque was also found ($X^2 = 17.728$; $p < 0.001$), indicating higher z-coherence values during the 50% MVC compared to the 25% MVC condition. The values at 25% and 50% MVC were 1.30 (0.45) and 1.44 (0.34), respectively. In addition, a significant Group x Torque interaction was detected ($X^2 = 8.148$; $p = 0.004$) (**Figure 7b**). Post-hoc analyses revealed a significant difference between groups at 25% MVC ($p < 0.001$), but not at 50% MVC ($p = 0.110$), indicating that older participants exhibited lower z-coherence values than younger participants only at the lower intensity level. Moreover, higher z-coherence values at 50% MVC compared to 25% MVC were observed in older participants ($p < 0.001$), but not in younger individuals ($p = 0.503$). Lastly, a significant Group x Torque x Muscle interaction was found ($X^2 = 4.093$; $p = 0.043$) (**Figure 7c**). Post-hoc comparisons showed a significant increase in z-coherence values from 25% to 50% MVC in the left LES of older participants ($p < 0.001$), whereas no change was observed in the right LES ($p = 1.000$). In younger participants, z-coherence values did not differ between contraction levels for either muscle ($p > 0.05$). In addition, a significant difference between groups was observed for left LES at 25% MVC ($p < 0.001$), with older participants showing lower z-coherence values. A trend toward a similar difference was found for the right LES ($p = 0.054$). Instead, no significant differences emerged at 50% MVC ($p > 0.05$).



(c)

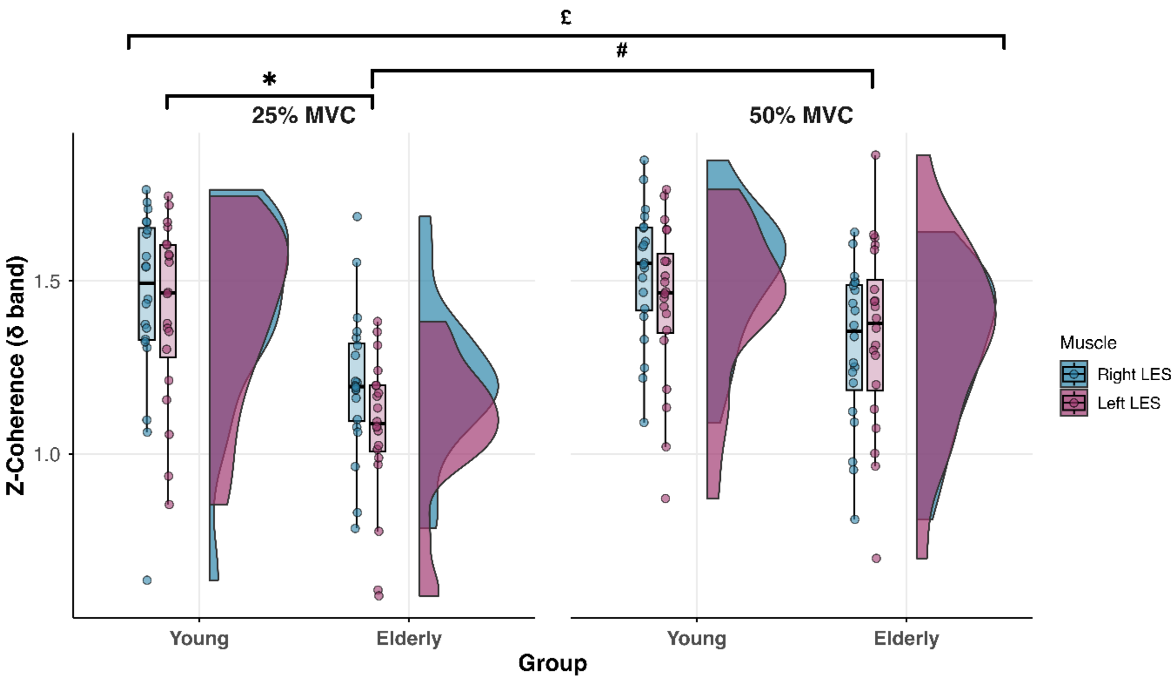


Figure 7. Overall z-coherence values **(a)**, z-coherence values at 25% and 50% of maximal voluntary contraction (MVC) **(b)**, and z-coherence values in the right and left lumbar erector spinae (LES) at each contraction intensity **(c)** in young and elderly participants during the isokinetic trunk extension contractions. Median (IQR). The distributions are illustrated using half-violin plots with embedded boxplots and individual data points; ***** Main effect of Group. **#** Main effect of Torque. **$** Group x Torque interaction. **£** Group x Torque x Muscle interaction.

4.5 Discussion

The present study investigated the relationship between LES HDsEMG activity and torque fluctuations during isometric and isokinetic concentric trunk extension contractions in young and older adults, using coherence analysis (0-5 Hz). In addition, regional differences in HDsEMG-torque coherence of the LES muscles were characterised in both groups. Consistent with our hypothesis, older adults exhibited greater torque CoV during both isometric and isokinetic contractions, indicating reduced control of trunk extensor muscle force. Notably, the age-related difference was larger during isokinetic contractions and, within these, was most pronounced at the lower intensity (25% MVC). Contrary to our expectations, the older group exhibited a weaker correlation between HDsEMG and torque (i.e., lower coherence values) only during isokinetic contractions, whereas alterations in the regional distribution of LES HDsEMG-torque coherence were evident exclusively during isometric contractions.

Age-related reductions in maximal torque

It is well-documented in the literature that isometric maximal strength/torque of the trunk extensor muscles declines with age (Singh et al. 2011, 2013; Porto et al. 2020; Parrella et al. 2025). In contrast, fewer studies have examined age-related differences in trunk strength during isokinetic contractions. Existing evidence indicates that older adults exhibit lower peak torque values than younger individuals across multiple angular velocities, ranging from slow (15°/s) to fast (180°/s) contraction speeds (Granito et al. 2014; Dallaway et al. 2021). Accordingly, the present study also observed reduced maximal isokinetic torque in older adults, assessed at a lower angular velocity (10°/s). Therefore, our findings extend previous evidence by demonstrating that the age-related decline in trunk extensor strength is also present under slower contraction conditions. A recent study has shown that this decline may not be primarily attributed to age-related atrophy and fat infiltration of the paravertebral muscles, but rather to neuromuscular mechanisms such as a decline in the neural drive to the muscles or an age-related remodeling of muscle phenotype (Dallaway et al. 2021).

Moreover, neither isometric nor isokinetic maximal muscle torque decreased following the respective submaximal contraction tasks in either group. This suggests that no significant muscle fatigue was induced during the tasks, which is important as fatigue could have represented a potential source of bias in both the mechanical (i.e., muscle force control) and physiological assessments (Pethick and Tallent 2022).

Impaired torque steadiness with ageing

The reduced torque steadiness (i.e., greater force fluctuations) observed in older individuals in the present study aligns with previous evidence of impaired muscle force control with ageing, both across various muscle groups (Pethick et al. 2022) and, although less frequently investigated, in the trunk extensors. In particular, previous studies have shown an age-related increase in force fluctuations of the trunk extensor muscles during a fatiguing task performed at 30% of maximal voluntary isometric force (Parrella et al. 2025) and during a 15-s steadiness task at 10% of peak torque (Porto et al. 2020). Similarly, trunk position and force sense have been shown to be altered in older individuals with hyperkyphosis (Keshavarzi et al. 2022). However, in contrast to the present findings, no differences in force fluctuations between young and older participants were reported at 50% of peak torque (Porto et al. 2020). This discrepancy may be explained by different factors, such as the younger mean age of the older group in that study (69.22 ± 4.10 years) compared with the present sample, as well as differences in the experimental setup. All the studies mentioned so far have focused on isometric contractions of the trunk extensor muscles, whereas muscle force control during dynamic contractions has been investigated only in individuals with chronic low back pain (Arvanitidis et al. 2024b). Thus, to the best of the authors' knowledge, the present study is the first to examine force control of the trunk extensor muscles during dynamic contractions in older adults.

Notably, we found that the difference in force steadiness between the two groups was greater during isokinetic contractions compared to isometric ones. This finding aligns with some previous studies reporting larger age-related increases in force fluctuations during sinusoidal, concentric, and eccentric tasks than during isometric contractions (Burnett et al. 2000; Keogh et al. 2006). Such differences may have meaningful functional implications, as dynamic contractions more closely reflect the movements performed during everyday activities. Moreover, during the isokinetic contractions, the difference in force steadiness between groups was more pronounced at 25% MVC than at 50% MVC, indicating that age-related impairments in force control were greater at lower contraction intensities. This finding is consistent with previous evidence showing reduced stability in older adults particularly at lower target forces, such as during slow anisometric contractions of the first dorsal interosseous muscle at lighter loads (Laidlaw et al. 2000) and during low-intensity isokinetic contractions of the quadriceps (Hortobágyi et al. 2001). This age-related increase in force fluctuations at predominantly lower contraction intensities is particularly relevant, as most activities of daily living, especially those commonly performed by older adults, typically require forces of up to approximately 20% MVC (Pethick et al. 2022).

Age-related alterations in HDsEMG-torque coherence

The results of the present study revealed significant differences in HDsEMG-torque coherence values between groups during the isokinetic trunk extension contractions, whereas no differences were observed during the isometric contractions. This finding is somewhat surprising, as we expected lower HDsEMG-torque coherence in older adults during both isometric and isokinetic tasks, given that they exhibited greater torque fluctuations than younger participants in both conditions. Indeed, lower coherence between HDsEMG activity and torque would reflect reduced coupling between the neural input to the motor-unit pool and the mechanical output, which can, in turn, alter force stability (Arvanitidis et al. 2024b).

The decrease in coherence between oscillations in LES muscle activity and torque generation observed in older adults during isokinetic contractions was primarily driven by a reduction in coherence at the lower contraction intensity (25% MVC). This reduction may reflect compensatory or altered neuromuscular strategies adopted by older adults, such as increased recruitment of synergistic muscles to generate torque, potentially contributing to greater force fluctuations. At lower intensities, which typically require greater precision and coordination (Porto et al. 2020), older adults may have reduced the contribution of the LES muscles to the resultant torque and redistributed the load across multiple muscles to a greater extent to generate the required torque. This may have led to increased force variability, as recruiting additional synergistic muscles can amplify force fluctuations due to each muscle operating along slightly different action directions (Kutch et al. 2008). Additionally, greater coactivation in older adults cannot be excluded, which may have further influenced coherence values and torque generation, as alternating activation of agonist and antagonist muscles can increase fluctuations in force and acceleration (Enoka et al. 2003). Importantly, the reduction in HDsEMG-torque coherence occurred at the contraction intensity where older adults exhibited the greatest impairment in force steadiness, with torque CoV being 107.84% higher compared to the younger group. In contrast, no differences in HDsEMG-torque coherence were observed at 50% MVC, although older adults still exhibited reduced force steadiness. At this higher intensity, it is plausible that older adults were required to activate the LES muscles more, contributing to torque generation similarly to younger individuals. Nevertheless, they may have continued to rely more on synergistic muscles than younger participants, thereby resulting in greater force fluctuations, although to a lesser extent than at 25% MVC. Altogether, these results suggest that previous evidence of age-related reductions in neural-mechanical coupling, particularly at lower contraction intensities (Laidlaw et al. 2000; Enoka et al. 2003), also applies to the trunk extensor muscles, at least during isokinetic contractions. Lastly, at 25% MVC, a difference between muscles was observed in older adults, with the left LES muscle contributing less to torque. However, as shown in Figure 7c, a similar trend was also observed for the right LES muscle. Therefore, future studies should confirm or refute this difference, as it may reflect an asymmetry in torque contribution between the LES muscles in older adults, potentially resulting from postural compensations associated with ageing. Besides age-related differences, a potential mechanical contributor to force fluctuations in both groups is the effect of muscle-tendon length on force/torque production capacity (Bampouras et al., 2017; Kellis and Blazevich, 2022). In particular, torque production in the hip and trunk extensors varies with hip angle due to the force-length relationship (Raschke and Chaffin 1996; Kellis and Blazevich, 2022). Therefore, variations in muscle-tendon length may have contributed to the challenge of controlling torque during the isokinetic contractions.

In contrast to dynamic contractions, no differences in HDsEMG-torque coherence values between groups were observed during the isometric trunk extension contractions. During these contractions, it is possible that older adults were unable to use the compensatory strategies likely adopted during the isokinetic contractions due to the more constrained position. Nevertheless, they were still less stable than the younger individuals. As reported by Enoka et al. (2003), during simpler tasks such as isometric contractions, the properties of individual motor units within a single muscle play a more dominant role in force control, whereas in more complex movements (e.g., isokinetic contractions), force fluctuations depend more on the distribution of activity among multiple muscles. Therefore, it is possible that, although the LES muscles contribute similarly to torque generation in both young and older individuals, inherent alterations in the common synaptic input to the motor unit pool (i.e., in the oscillatory components of this input) in older adults may underlie their reduced force steadiness (Castronovo et al. 2018). However, to study these mechanisms, motor unit decomposition analysis would be necessary. Unfortunately, this remains challenging in the trunk muscles due to several factors, such as the complex spinal anatomy (e.g., the presence of multiple muscle layers and thoracolumbar fascia) and the volume conductor characteristics of the lower lumbar region (e.g., thick subcutaneous tissue layer) (Christophy et al. 2012). Although HDsEMG-torque coherence values were similar between groups, differences emerged in their spatial distribution, with older adults showing a more cranial and medial coherence pattern within the LES muscles. The greater contribution to torque from the more cranial regions in older adults is consistent with our previous study (Parrella et al. 2025) and may reflect a compensatory mechanism aimed at redistributing the load toward the upper lumbar regions, as the L4-L5 segments commonly exhibit greater degeneration with ageing (Hicks et al.

2009). Instead, the more medial coherence may be related to greater recruitment of the superficial multifidus muscles located closer to the spine, possibly to help stabilise the trunk, particularly in the presence of the more flexed posture commonly associated with ageing (Abboud et al. 2023). This neuromuscular strategy adopted by older individuals may contribute to their reduced torque steadiness, as a more flexed posture and greater recruitment of the upper lumbar vertebrae can limit proper activation of the LES muscles (Arvanitidis et al. 2024b).

Age-related differences in RMSnorm

The results of the present study demonstrated overall greater muscle activation (i.e., higher RMSnorm values) in older adults during both isometric and isokinetic submaximal contractions. This finding is consistent with previous studies reporting increased muscle activation in older adults compared to younger individuals to generate the same amount of force during force steadiness tasks, along with greater force fluctuations (Laidlaw et al. 2000; Tracy 2007; Keenan and Massey 2012). However, during the isokinetic trunk extensions, this difference in muscle activation was primarily driven by greater activation of the right LES muscle in the older group, whereas no such difference between muscles was found during isometric contractions. Although RMS and coherence analyses assess different aspects of neuromuscular function, the observed difference in activation between muscles further supports the presence of potential asymmetries between the LES muscles in older adults during dynamic contractions. More specifically, this finding may reflect postural adaptations associated with ageing that are not evident during isometric contractions but become apparent during dynamic tasks, which better reveal underlying motor impairments. In line with this interpretation, some previous studies have reported age-related deficits or asymmetries emerging primarily during dynamic or challenging tasks, while remaining small or absent during static or isometric tests (Baloh et al. 1994; Hill et al. 2020; Heap-Eldridge et al. 2024).

Methodological considerations and limitations

A limitation of this study is that the sample consisted of healthy, well-functioning older adults, which limits the generalisability of the findings to other populations, such as frail older individuals. However, as this is among the few studies to have examined force control of the trunk extensor muscles in the context of ageing, it provides a good basis for future research involving populations at greater risk. In addition, the present study did not assess potential compensatory strategies, such as the activation of synergistic muscles (e.g., hip extensors) or antagonist co-activation, which may have contributed to task performance. Lastly, we acknowledge that the rectified sEMG provides only an indirect and coarse estimate of the neural drive to the muscles, as it can be influenced by several physiological and methodological factors. A more accurate estimation of the neural drive can be achieved using HDsEMG decomposition analysis, which allows the identification of specific motor unit discharge patterns. However, as previously mentioned, this approach remains challenging in the trunk muscles.

4.6 Conclusions

The present study showed that force control of the trunk extensor muscles is impaired in older individuals. Specifically, we uniquely demonstrated that age-related impairments in torque steadiness are more pronounced during isokinetic than isometric trunk extension contractions, with the greatest deficits observed at the lower contraction intensity (25% MVC). Moreover, distinct neuromuscular patterns emerged depending on contraction type: older adults exhibited reduced HDsEMG-torque coherence magnitude during isokinetic contractions and an altered spatial distribution of coherence during isometric contractions. Overall, the present study contributes to the existing body of knowledge on muscle force control (i.e., force/torque steadiness) in older adults and extends current understanding to the trunk extensor muscles.

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CHAPTER 5

General discussion

5.1 Summary of findings

The main aim of this PhD thesis was to enhance the understanding of age-related adaptations of the trunk extensor muscles, particularly in relation to endurance capacity, force control (torque steadiness), and the potential underlying neuromuscular mechanisms. The main findings of the three studies can be summarised as follows:

1) The first study (**Chapter 2**) demonstrated that ageing influences the neuromuscular adaptations of the LES muscle to fatigue. Specifically, both older and younger individuals exhibited a redistribution of LES activity in response to fatigue, which may explain the similar endurance times observed between groups, as such redistribution likely serves to prolong contraction (Farina et al. 2008). However, the patterns of redistribution differed between groups. Older adults exhibited a cranial shift in the centroid of LES muscle activity, whereas younger individuals displayed comparable activation of both caudal and cranial regions, accompanied by a lateral displacement of the centroid with fatigue. The distinct activation patterns observed in older adults may represent a compensatory neuromuscular strategy aimed at reducing compressive and shear forces on the lower lumbar spine, particularly at the L4-L5 level, which is more prone to degeneration with ageing (Hicks et al., 2009). Nevertheless, this strategy may be biomechanically less efficient, as the load is distributed across fewer fibres and the cranial LES primarily extends the upper lumbar vertebrae rather than the entire lumbar spine (Sanderson et al., 2019). This may contribute to the greater force fluctuations observed in older adults during the fatiguing task.

2) The meta-analysis presented in **Chapter 3**, which included nine studies, demonstrated that endurance time of the back extensor muscles is reduced during isometric contractions in older adults. Although the studies employed various isometric protocols, the observed effect consistently indicated a decline in endurance with ageing. This age-related decline may be partly attributable to fat infiltration within the paraspinal muscles, as consistent evidence

indicates a negative correlation between fat infiltration and back muscle endurance (Yazici and Yerlikaya 2022; Han et al. 2024; Liu et al. 2024). However, from a neuromuscular perspective, although some studies included EMG assessments, variability in outcomes and methodological approaches across studies limits the ability to draw definitive conclusions.

3) The third study (**Chapter 4**) demonstrated that force control of the trunk extensor muscles is impaired in older individuals. Specifically, age-related impairments in torque steadiness were more pronounced during isokinetic than isometric trunk extension contractions, particularly at the lower intensity (25% MVC). This finding is relevant as most activities of daily living performed by older adults typically require forces up to approximately 20% MVC (Pethick et al. 2022). Moreover, distinct neuromuscular patterns emerged depending on contraction type. During isokinetic contractions, older adults showed reduced HDsEMG-torque coherence, possibly reflecting altered neuromuscular strategies, such as increased recruitment of synergistic muscles or greater coactivation to generate torque. In contrast, during isometric contractions, although overall coherence values did not differ between groups, older adults exhibited higher HDsEMG-torque coherence in more cranial and medial regions of the LES muscles. This may indicate compensatory mechanisms to redistribute load toward the upper lumbar regions while attempting to enhance trunk stability through greater recruitment of the superficial multifidus. Overall, these altered neuromuscular patterns may limit optimal force production in older adults.

5.2 General conclusion

Taken together, this PhD thesis provides novel insights into how ageing affects the neuromuscular function of the trunk extensor muscles.

The findings demonstrate that the neuromuscular activation patterns of the LES muscle during fatigue are altered in older individuals. Although this may represent a compensatory adaptive response to age-related degenerative changes in the spine, it could also negatively affect physical function in older adults over the long term. Indeed, such strategy may lead to increased tissue loading in specific regions of the muscle and may potentially contribute to the development of musculoskeletal conditions such as low back pain, a condition that is highly prevalent with ageing (de Souza et al. 2019). Based on the results of the meta-analysis, this thesis also demonstrates that ageing is associated with a decline in the endurance of the back extensor muscles during isometric tasks. This finding is particularly interesting, as it contrasts with the so-called "fatigue paradox" reported in older individuals for other muscles, such as the elbow flexors and knee extensors, which suggests an increase in fatigue resistance with ageing (Christie et al. 2011). While differences in muscle-specific properties of the trunk extensor muscles are likely a primary contributor to this discrepancy, differences in the task characteristics used to assess endurance may also partly explain it. Trunk extensor muscle endurance is a key determinant of functional ability (Suri et al. 2011; Mesquita et al. 2019) and a predictor of participation in life roles in older adults (Beauchamp et al. 2016). Therefore, this parameter should be specifically considered when designing training interventions for this population. Moreover, the results of the present thesis show that older adults exhibit impaired force control (torque steadiness) of the trunk extensor muscles during both isometric and dynamic contractions, particularly during low-intensity dynamic trunk extensions. These impairments are accompanied by alterations in neuromuscular recruitment strategies of the LES muscles. Overall, these findings support the notion that force/torque steadiness is impaired with ageing and that age-related motor adaptations are task-dependent, as previously demonstrated in other muscles (Pethick et al. 2022). In particular, age-related adaptations may be more pronounced during challenging tasks, such as dynamic movements that require higher mechanical demands and greater neuromuscular coordination. This may lead older individuals to rely on more compensatory and variable movement strategies, which could ultimately result in greater impairments in motor control.

Collectively, this thesis provides evidence that ageing negatively affects motor performance of the trunk extensor muscles and reveals significant differences in muscle activation patterns between older and younger adults. These findings enhance the understanding of neuromuscular adaptations in the LES muscles with ageing, which is particularly important given that the paraspinal muscles are postural muscles essential for maintaining upright posture and performing basic daily activities.

5.3 Future directions

The findings of this thesis pave the way for future research in older individuals. Potential future directions include:

- Further assessment of dynamic trunk movements, including muscle endurance and fatigue patterns during dynamic trunk extensions, as well as muscle force control during eccentric trunk extension contractions.
- Incorporation of cognitive tasks during both fatiguing and force steadiness assessments, as previously applied in other muscles (Hunter et al. 2016), to better simulate the complexity of activities of daily living.
- Investigation of the effects of trunk extensor muscle fatigue on postural stability during functional tasks.
- Combined assessment of muscle force control and fatigue tasks, for instance by assessing force steadiness pre- and post-fatigue to examine how fatigue affects it, providing a more comprehensive understanding of age-related neuromuscular alterations.
- Comparison of healthy older adults with those experiencing low back pain, assessing both motor performance (endurance and force control) of the trunk extensor muscles and the associated neuromuscular patterns, given the high prevalence and disabling impact of low back pain in ageing populations (de Souza et al. 2019).

- Development of longitudinal training studies targeting endurance capacity and torque steadiness of the trunk extensor muscles in older adults, potentially integrating advanced technologies such as HDsEMG-based biofeedback to restore optimal motor function.

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APPENDIX

Supplemental files – Chapter 3

Supplemental file 1 – PRISMA Checklist

Section and Topic	Item #	Checklist item	Location where item is reported
TITLE			
Title	1	Identify the report as a systematic review.	Title
ABSTRACT			
Abstract	2	See the PRISMA 2020 for Abstracts checklist.	Abstract
INTRODUCTION			
Rationale	3	Describe the rationale for the review in the context of existing knowledge.	Introduction (central part)
Objectives	4	Provide an explicit statement of the objective(s) or question(s) the review addresses.	Introduction (last few lines)
METHODS			
Eligibility criteria	5	Specify the inclusion and exclusion criteria for the review and how studies were grouped for the syntheses.	Eligibility criteria; synthesis methods
Information sources	6	Specify all databases, registers, websites, organisations, reference lists and other sources searched or consulted to identify studies. Specify the date when each source was last searched or consulted.	Information sources
Search strategy	7	Present the full search strategies for all databases, registers and websites, including any filters and limits used.	Supplemental file 2
Selection process	8	Specify the methods used to decide whether a study met the inclusion criteria of the review, including how many reviewers screened each record and each report retrieved, whether they worked independently, and if applicable, details of automation tools used in the process.	Selection process
Data collection process	9	Specify the methods used to collect data from reports, including how many reviewers collected data from each report, whether they worked independently, any processes for obtaining or confirming data from study investigators, and if applicable, details of automation tools used in the process.	Data collection process and data items
Data items	10a	List and define all outcomes for which data were sought. Specify whether all results that were compatible with each outcome domain in each study were sought (e.g. for all measures, time points, analyses), and if not, the methods used to decide which results to collect.	Outcomes; data collection process and data items
	10b	List and define all other variables for which data were sought (e.g. participant and intervention characteristics, funding sources). Describe any assumptions made about any missing or unclear information.	Data collection process and data items
Study risk of bias assessment	11	Specify the methods used to assess risk of bias in the included studies, including details of the tool(s) used, how many reviewers assessed each study and whether they worked independently, and if applicable, details of automation tools used in the process.	Risk of bias assessment
Effect measures	12	Specify for each outcome the effect measure(s) (e.g. risk ratio, mean difference) used in the synthesis or presentation of results.	Synthesis

Section and Topic	Item #	Checklist item	Location where item is reported
			methods
Synthesis methods	13a	Describe the processes used to decide which studies were eligible for each synthesis (e.g. tabulating the study intervention characteristics and comparing against the planned groups for each synthesis (item #5)).	Synthesis methods
	13b	Describe any methods required to prepare the data for presentation or synthesis, such as handling of missing summary statistics, or data conversions.	Synthesis methods
	13c	Describe any methods used to tabulate or visually display results of individual studies and syntheses.	Synthesis methods
	13d	Describe any methods used to synthesize results and provide a rationale for the choice(s). If meta-analysis was performed, describe the model(s), method(s) to identify the presence and extent of statistical heterogeneity, and software package(s) used.	Synthesis methods
	13e	Describe any methods used to explore possible causes of heterogeneity among study results (e.g. subgroup analysis, meta-regression).	Synthesis methods; sensitivity analyses
	13f	Describe any sensitivity analyses conducted to assess robustness of the synthesized results.	Sensitivity analyses
Reporting bias assessment	14	Describe any methods used to assess risk of bias due to missing results in a synthesis (arising from reporting biases).	Certainty of evidence
Certainty assessment	15	Describe any methods used to assess certainty (or confidence) in the body of evidence for an outcome.	Certainty of evidence
RESULTS			
Study selection	16a	Describe the results of the search and selection process, from the number of records identified in the search to the number of studies included in the review, ideally using a flow diagram.	Study selection; Figure 1
	16b	Cite studies that might appear to meet the inclusion criteria, but which were excluded, and explain why they were excluded.	Study selection
Study characteristics	17	Cite each included study and present its characteristics.	Study selection
Risk of bias in studies	18	Present assessments of risk of bias for each included study.	Risk of bias assessment; Table 2; Supplemental file 3
Results of individual studies	19	For all outcomes, present, for each study: (a) summary statistics for each group (where appropriate) and (b) an effect estimate and its precision (e.g. confidence/credible interval), ideally using structured tables or plots.	Study characteristics; Table 1
Results of syntheses	20a	For each synthesis, briefly summarise the characteristics and risk of bias among contributing studies.	Table 2; Supplemental file 3

Section and Topic	Item #	Checklist item	Location where item is reported
	20b	Present results of all statistical syntheses conducted. If meta-analysis was done, present for each the summary estimate and its precision (e.g. confidence/credible interval) and measures of statistical heterogeneity. If comparing groups, describe the direction of the effect.	Meta-analysis results for endurance time; Figure 2
	20c	Present results of all investigations of possible causes of heterogeneity among study results.	Refined analysis after outlier removal; Figure 3; Supplemental File 7
	20d	Present results of all sensitivity analyses conducted to assess the robustness of the synthesized results.	Refined analysis after outlier removal; Figure 4
Reporting biases	21	Present assessments of risk of bias due to missing results (arising from reporting biases) for each synthesis assessed.	Assessment of publication bias; Figure 5
Certainty of evidence	22	Present assessments of certainty (or confidence) in the body of evidence for each outcome assessed.	GRADE; Table 3
DISCUSSION			
Discussion	23a	Provide a general interpretation of the results in the context of other evidence.	Discussion (central part)
	23b	Discuss any limitations of the evidence included in the review.	Discussion (methodological considerations)
	23c	Discuss any limitations of the review processes used.	Discussion (methodological considerations)
	23d	Discuss implications of the results for practice, policy, and future research.	Discussion (methodological considerations, conclusions)
OTHER INFORMATION			
Registration and protocol	24a	Provide registration information for the review, including register name and registration number, or state that the review was not registered.	Methods (first few lines)
	24b	Indicate where the review protocol can be accessed, or state that a protocol was not prepared.	Methods (first few lines)

Section and Topic	Item #	Checklist item	Location where item is reported
	24c	Describe and explain any amendments to information provided at registration or in the protocol.	Not applicable
Support	25	Describe sources of financial or non-financial support for the review, and the role of the funders or sponsors in the review.	Declaration of funding sources
Competing interests	26	Declare any competing interests of review authors.	Conflict of interest
Availability of data, code and other materials	27	Report which of the following are publicly available and where they can be found: template data collection forms; data extracted from included studies; data used for all analyses; analytic code; any other materials used in the review.	Data availability statement

From: Page MJ, McKenzie JE, Bossuyt PM, Boutron I, Hoffmann TC, Mulrow CD, et al. The PRISMA 2020 statement: an updated guideline for reporting systematic reviews. BMJ 2021;372:n71. doi: 10.1136/bmj.n71

For more information, visit: http://www.prisma-statement.org/

Supplemental file 2 – Search strategy for all databases

- **<u>SEARCH STRATEGY</u>** (Medline, EMBASE, PubMed, Web of Science, CINAHL)

MEDLINE / EMBASE

1) Spine/

2) Spinal muscles.mp

3) Back Muscles/

4) Paraspinal Muscles/

5) Erector spinae.mp.

6) Multifidus.mp.

7) Trunk extensor muscles.mp.

8) Lumbar extensor muscles.mp.

9) Neck muscles/

10) Neck extensor muscles.mp.

11) Cervical extensor muscles.mp.

12) Splenius muscles.mp.

13) 1 – 12 (OR)

14) Muscle Fatigue/

15) Fatigue/

16) Neuromuscular fatigue.mp.

17) Fatigability.mp.

18) Endurance.mp.

19) Exercise tolerance/

20) Task failure.mp.

21) 14 – 20 (OR)

22) Aging/

23) Healthy Aging/

24) Ageing.mp.

25) Aged/

26) Older people.mp.

27) Older adults.mp.

28) Elderly.mp.

29) Elders.mp.

30) 22 – 29 (OR)

31) **13 AND 21 AND 30 - <u>TOT</u>: 83 papers for MEDLINE – 253 papers for EMBASE**

PUBMED

(("Spine"[MeSH] OR "Spinal muscles" OR "Back Muscles"[MeSH] OR "Paraspinal Muscles"[MeSH] OR "Erector Spinae" OR "Multifidus" OR "Trunk Extensor Muscles" OR "Lumbar Extensor Muscles" OR "Neck Muscles"[MeSH] OR "Neck Extensor Muscles" OR "Cervical Extensor Muscles" OR "Splenius Muscles") AND ("Muscle Fatigue"[MeSH] OR "Fatigue"[MeSH] OR "Neuromuscular Fatigue" OR "Fatigability" OR "Endurance" OR "Exercise Tolerance"[MeSH] OR "Task Failure") AND ("Aging"[MeSH] OR "Healthy Aging"[MeSH] OR "Aged"[MeSH] OR "Older People" OR "Older Adults" OR "Elderly" OR "Elders" OR "Ageing")) - **<u>TOT</u>: 154 papers**

Web of science

https://www.webofscience.com/wos/alldb/summary/edd1e4a9-175e-47c4-be0c-32deac02b1b9-0147ab5481/relevance/1

(TS=("spine" OR "spinal muscles" OR "back muscles" OR "paraspinal muscles" OR "neck muscles"
OR "erector spinae" OR "multifidus" OR "trunk extensor muscles"
OR "lumbar extensor muscles" OR "neck extensor muscles"
OR "cervical extensor muscles" OR "splenius muscles"))
AND
(TS=("muscle fatigue" OR "fatigue" OR "neuromuscular fatigue" OR "fatigability"
OR "endurance" OR "exercise tolerance" OR "task failure"))
AND
(TS=("aging" OR "healthy aging" OR "aged" OR "older people"
OR "older adults" OR "elderly" OR "elders" OR "ageing")) and Preprint Citation Index (Exclude – Database) and Other or Review Article (Exclude – Document Types) and Dissertation Thesis or Awarded Grant or Data Set or Retracted Publication or Unspecified (Exclude – Document Types) and Patent (Exclude – Document Types) and Book (Exclude – Document Types)

<u>TOT</u>: 734 papers

CINAHL Plus

1) (MH "Spine")

2) "Spinal muscles"

3) "back muscles"

4) "Paraspinal Muscles"

5) (MH "Erector Spinae Muscles") OR (MH "Multifidus Muscles")

6) "Trunk Extensor Muscles"

7) "Lumbar Extensor Muscles"

8) (MH "Neck Muscles")

9) "Neck Extensor Muscles"

10) "Cervical Extensor Muscles"

11) "Splenius Muscles"

12) (MH "Muscle Fatigue")

13) (MH "Fatigue")

14) "neuromuscular fatigue"

15) "Fatigability"

16) "endurance"

17) (MH "Exercise Tolerance")

18) "task failure"

19) (MH "Aging") OR (MH "Healthy Aging")

20) (MH "Aged")

21) "Older People"

22) "Older Adults"

23) "Elderly"

24) "Elders"

25) "Ageing"

26) 1 OR 2 OR 3 OR 4 OR 5 OR 6 OR 7 OR 8 OR 9 OR 10 OR 11

27) 12 OR 13 OR 14 OR 15 OR 16 OR 17 OR 18

28) 19 OR 20 OR 21 OR 22 OR 23 OR 24 OR 25

29) 26 AND 27 AND 28

TOT: 29 papers

TOT N PAPERS: 1253

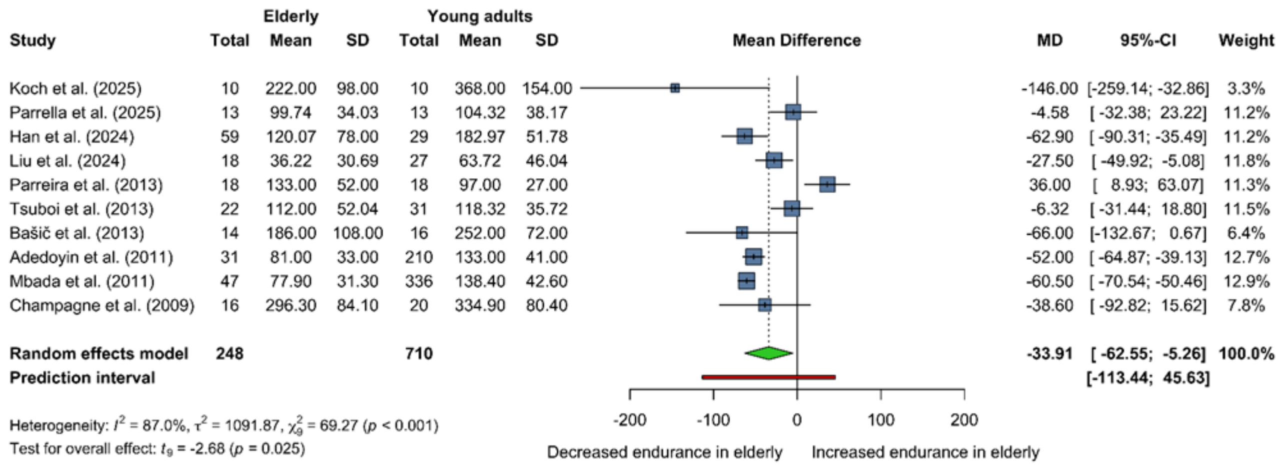
Supplemental file 3 – Full set of questions for risk of bias assessment

	Koch et al. (2025)	Parrella et al. (2025)	Han et al. (2024)	Liu et al. (2024)	Da Silva et al. (2015)	Kienbacher et al. (2014)	Parreira et al. (2013)	Tsuboi et al. (2013)	Bašič et al. (2013)	Adedoyin et al. (2011)	Mbada et al. (2011)	Singh et al. (2011)	Champagne et al. (2009)
Introduction													
1. Were the aims/objectives of the study clear?	*Yes*	*Yes*	*Yes*	*Yes*	*Yes*	*Yes*	*Yes*	*Yes*	*Yes*	*Yes*	*Yes*	*Yes*	*Yes*
Methods													
2. Was the study design appropriate for the stated aim(s)?	*Yes*	*Yes*	*Yes*	*Yes*	*Yes*	*Yes*	*Yes*	*Yes*	*Yes*	*Yes*	*Yes*	*Yes*	*Yes*
3. Was the sample size justified?	*Yes*	*Yes*	*No*	*No*	*Yes*	*No*	*No*	*No*	*No*	*No*	*No*	*No*	*No*
4. Was the target/reference population clearly defined? (Is it clear who the research was about?)	*Yes*	*Yes*	*Yes*	*Yes*	*Yes*	*Yes*	*Yes*	*Yes*	*Yes*	*Yes*	*Yes*	*Yes*	*Yes*
5. Was the sample frame taken from an appropriate population base so that it closely represented the target/reference population under investigation?	*Yes*	*No*	*Yes*	*Don't know*	*No*	*Yes*	*No*	*Don't know*	*Don't know*	*Yes*	*Yes*	*Yes*	*Yes*
6. Was the selection process likely to select subjects/participants that	*Yes*	*Yes*	*Yes*	*Yes*	*Yes*	*Yes*	*Yes*	*No*	*No*	*Yes*	*Yes*	*Yes*	*Yes*

were representative of the target/reference population under investigation?													
7. Were the risk factor and outcome variables measured appropriate to the aims of the study?	*Yes*	*Yes*	*Yes*	*Yes*	*Yes*	*Yes*	*Yes*	*Yes*	*Yes*	*Yes*	*Yes*	*Yes*	*Yes*
8. Were the risk factor and outcome variables measured correctly using instruments/measurements that had been trialled, piloted or published previously?	*Yes*	*Yes*	*Yes*	*Yes*	*No*	*No*	*Yes*	*Yes*	*Yes*	*Yes*	*Yes*	*Yes*	*No*
9. Is it clear what was used to determined statistical significance and/or precision estimates? (e.g. p-values, confidence intervals)	*Yes*	*Yes*	*Yes*	*Yes*	*Yes*	*No*	*Yes*	*Yes*	*Yes*	*Yes*	*Yes*	*Yes*	*Yes*
10. Were the methods (including statistical methods) sufficiently described to enable them to be repeated?	*Yes*	*Yes*	*Yes*	*Yes*	*Yes*	*No*	*Yes*	*Yes*	*Yes*	*No*	*No*	*Yes*	*Yes*
Results													
11. Were the basic data adequately described?	*No*	*Yes*	*No*	*No*	*No*	*No*	*Yes*	*No*	*Yes*	*No*	*No*	*No*	*Yes*
12. Was information on participant dropouts during testing (e.g., rates and reasons) reported?	*Yes*	*No*	*No*	*No*	*No*	*Yes*	*No*	*No*	*No*	*No*	*No*	*No*	*Yes*

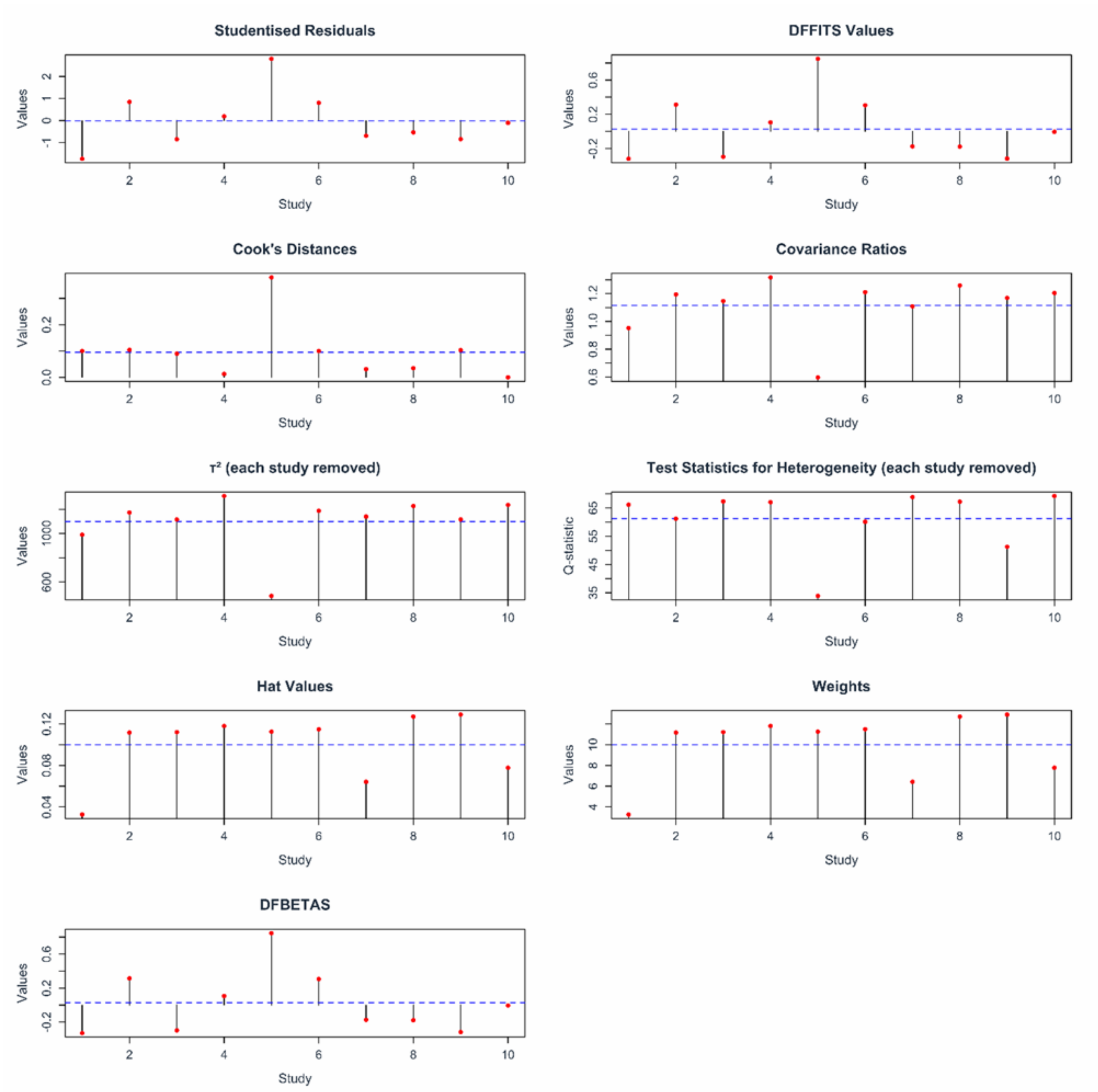
13. Were the results internally consistent?	*Yes*	*Yes*	*Yes*	*No*	*Yes*	*Yes*	*No*	*Yes*	*Yes*	*Yes*	*Yes*	*No*	*Yes*
14. Were the results presented for all the analyses described in the methods?	*Yes*	*Yes*	*Yes*	*Yes*	*Yes*	*Yes*	*Yes*	*Yes*	*No*	*Yes*	*Yes*	*Yes*	*Yes*
Discussion													
15. Were the authors' discussions and conclusions justified by the results?	*Yes*	*Yes*	*Yes*	*No*	*Yes*	*Yes*	*Yes*	*Yes*	*No*	*Yes*	*Yes*	*Yes*	*Yes*
16. Were the limitations of the study discussed?	*Yes*	*Yes*	*Yes*	*Yes*	*Yes*	*Yes*	*Yes*	*No*	*No*	*Yes*	*Yes*	*No*	*Yes*
Other													
17. Were there any funding sources or conflicts of interest that may affect the authors' interpretation of the results?	*No*	*No*	*No*	*No*	*No*	*No*	*No*	*Don't know*	*Don't know*	*Don't know*	*Don't know*	*Don't know*	*No*
18. Was ethical approval or consent of participants attained?	*Yes*	*Yes*	*Yes*	*Yes*	*Yes*	*Yes*	*Yes*	*Yes*	*Yes*	*Yes*	*Yes*	*Yes*	*Yes*
SCORE (/18)	**17**	**16**	**15**	**12**	**14**	**13**	**14**	**11**	**10**	**13**	**13**	**12**	**16**
SCORE (%)	**94,44**	**88,89**	**83,33**	**66,67**	**77,78**	**72,22**	**77,78**	**61,11**	**55,56**	**72,22**	**72,22**	**66,67**	**88,89**

Supplemental file 4 – Initial meta-analysis including all 10 studies



Supplemental file 4. Forest plot of endurance time differences between elderly and young adults including all 10 initially identified studies. The plot displays mean differences (MD) in endurance time (seconds) with corresponding 95% confidence intervals (95% CI) for each included study. The green diamond represents the pooled effect estimate with its 95% CI, while the red line indicates the prediction interval. Negative values on the x-axis indicate decreased endurance time in elderly compared to young adults, while positive values indicate increased endurance in elderly. The size of each square is proportional to the study's weight in the meta-analysis. The forest plot is organised in descending order based on publication year.

Supplemental file 5 – Initial influence diagnostics including all 10 studies



Supplemental file 5. Influence diagnostic plots for the meta-analysis of endurance time differences between elderly and young adults including all 10 initially identified studies. The figure presents an influence analysis for the meta-analytic model, illustrating various diagnostic measures to identify influential studies. The plots include externally studentised residuals, DFFITS values, Cook's distances, covariance ratios, estimates of heterogeneity (τ^2 & Q-statistic) when each study is removed, hat values, weights and DFBETAS. The y-axis shows the values of each measure, and the x-axis displays the study numbers, ordered according to their appearance in the forest plot. The blue line in each plot represents the mean of each parameter across all studies.

Supplemental file 6 – Specific values from the initial influence diagnostic analysis

Study	Is_Outlier	rstudent	dffits	cook_d	cov_r	tau2_del	QE_del	hat	weight	dfbetas
Koch et al. (2025)	FALSE	-1,735485822	-0,320353777	0,100270335	0,952471809	990,2988604	66,12105925	0,032626421	3,262642149	-0,32957874
Parrella et al. (2025)	FALSE	0,844221969	0,313372904	0,104417592	1,194728334	1173,69885	61,18122567	0,111626249	11,16262494	0,313624923
Han et al. (2024)	FALSE	-0,848158572	-0,29682178	0,089731791	1,146394475	1115,665445	67,29336933	0,112110087	11,21100872	-0,296901167
Liu et al. (2024)	FALSE	0,192753299	0,105661178	0,01315425	1,316858034	1309,805326	67,04942954	0,118046096	11,8046096	0,106334462
Parreira et al. (2013)	TRUE	2,800619951	0,84853343	0,379371327	0,596767186	485,0713318	33,86774611	0,112534066	11,2534066	0,846602804
Tsuboi et al. (2013)	FALSE	0,803466093	0,305218323	0,10023806	1,21034011	1187,329445	60,10764651	0,11490505	11,49050501	0,305814394
Bašič et al. (2013)	FALSE	-0,690222303	-0,174631705	0,031149955	1,108491843	1140,076803	68,83577961	0,064177114	6,41771138	-0,173373481
Adedoyin et al. (2011)	FALSE	-0,533907531	-0,177039071	0,035076171	1,259194385	1227,059896	67,192733	0,127167299	12,71672994	-0,178442777
Mbada et al. (2011)	FALSE	-0,840829236	-0,317896721	0,103326507	1,169345129	1116,961838	51,26196483	0,129088067	12,90880672	-0,318446441
Champagne et al. (2009)	FALSE	-0,102035564	-0,005185162	2,89732E-05	1,2045816	1236,044644	69,22930102	0,077719549	7,77195494	-0,00511293

Supplemental file 7 – Specific values from the secondary influence diagnostic analysis

Study	Is_Outlier	rstudent	dffits	cook_d	cov_r	tau2_del	QE_del	hat	weight	dfbetas
Koch et al. (2025)	FALSE	-1,72143	-0,25769	0,065854	0,972237	453,316	31,01015	0,022566	2,256556	-0,26312
Parrella et al. (2025)	FALSE	1,644643	0,588732	0,284193	0,921758	361,505	23,65979	0,125519	12,55192	0,593146
Han et al. (2024)	FALSE	-0,87533	-0,33023	0,11134	1,16999	499,3339	32,75561	0,126548	12,65483	-0,3301
Liu et al. (2024)	FALSE	0,566243	0,248043	0,071138	1,330274	581,2993	30,14333	0,13985	13,98502	0,24916
Tsuboi et al. (2013)	FALSE	1,642799	0,614441	0,300169	0,913526	352,0045	22,17776	0,132651	13,26515	0,615356
Bašič et al. (2013)	FALSE	-0,6197	-0,14314	0,020836	1,101639	512,8505	33,60316	0,05245	5,245013	-0,14138
Adedoyin et al. (2011)	FALSE	-0,45353	-0,16983	0,034477	1,372932	588,2629	33,50089	0,16307	16,30699	-0,17267
Mbada et al. (2011)	FALSE	-0,93971	-0,4265	0,178943	1,188125	476,7617	24,29895	0,168456	16,84559	-0,42574
Champagne et al. (2009)	FALSE	0,08175	0,039587	0,001654	1,191373	554,4168	33,73538	0,068889	6,88893	0,038682

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