UNIVERSITY OF VERONA

DEPARTMENT OF

Neurosciences, Biomedicine, and Movement Sciences

DOCTORAL PROGRAM IN

Neuroscience, Psychological and Psychiatric Sciences, and Movement Sciences

WITH THE FINANCIAL CONTRIBUTION OF

Microgate S.R.L., Bozen

CYCLE XXXVIII, year 2022

TITLE OF THE DOCTORAL THESIS

Gait Training con Treadmill Intelligente per il Parkinson (GaIT-PD)

Gait Training with Intelligent Treadmill for Parkinson's Disease (GaIT-PD)

S.S.D. Area 5, sector BIO-09

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La borsa di dottorato è stata cofinanziata con le risorse del PNRR:

Per il DM 352, nell'ambito della Missione 4 ("Istruzione e Ricerca") - Componente 2 ("Dalla Ricerca all'Impresa"), Investimento 3.3 ("Introduzione di dottorati innovativi che rispondono ai fabbisogni di innovazione delle imprese e promuovono l'assunzione dei ricercatori da parte delle imprese") - progetto M4C2 Investimento 3.3

SOMMARIO

La Malattia di Parkinson (MP) è un disturbo neurodegenerativo progressivo caratterizzato dalla degenerazione delle cellule dopaminergiche della substantia nigra, che determina bradicinesia, rigidità, tremore e importanti alterazioni dell'andatura (Bloem et al., 2021; Tolosa et al., 2006). Le modificazioni spazio-temporali del cammino, tra cui la riduzione della lunghezza del passo, l'aumento della cadenza, la diminuzione della velocità e l'incremento del tempo di doppio appoggio, rappresentano alcuni dei principali fattori che compromettono la mobilità e l'autonomia funzionale nella MP (Abbruzzese et al., 2016; Tomlinson et al., 2014). Nella MP, l'automatizzazione del cammino è ridotta a causa di un deficit nei processi di apprendimento motorio e di controllo basato sul feedback sensoriale, con un conseguente aumento della dipendenza da meccanismi di controllo anticipatorio (feedforward) (Nieuwboer et al., 2009; Olson et al., 2019). Questo deficit diventa particolarmente evidente nelle condizioni di dual-task (DT), in cui l'esecuzione simultanea di un compito cognitivo (come conteggio o sottrazioni in serie) provoca un ulteriore deterioramento dei parametri spazio-temporali dell'andatura (Kelly et al., 2012; Rochester et al., 2005). La valutazione del cammino in DT rappresenta quindi un indicatore sensibile della compromissione dell'automatizzazione locomotoria. Studi recenti hanno mostrato che i test di cammino in single e DT sono in grado di rilevare deficit locomotori già nelle fasi iniziali di malattia e di distinguere i pazienti con MP dai coetanei sani, suggerendo un ruolo diagnostico e prognostico di queste misure (Caronni et al., 2025; Zhang et al., 2022).

Parallelamente, un crescente corpo di evidenze indica che il cammino in DT è anche uno strumento utile per indagare gli effetti di interventi riabilitativi specifici. È importante distinguere tra due modalità di esercizio che, pur coinvolgendo componenti cognitive, agiscono su meccanismi differenti: DT training e Biofeedback (BF) training. Il DT training combina il cammino con un compito cognitivo aggiuntivo (ad esempio, conteggio, sottrazioni, flessibilità verbale). In questo tipo di intervento, le stesse funzioni cognitive misurate nei test di DT, attenzione divisa, memoria di lavoro, funzioni esecutive vengono direttamente allenate. Revisioni sistematiche e trial randomizzati dimostrano che questo approccio può migliorare la performance in DT, ridurre il DT cost e aumentare la velocità, la lunghezza del passo e la stabilità del cammino in persone con MP (Beck et al., 2018; Johansson et al., 2022; Sarasso et al., 2024; Wollesen et al., 2021; Yang et al., 2019; Zheng et al., 2021).

Il BF training, invece, non prevede l'esecuzione di un compito cognitivo separato, ma richiede un elevato livello di attenzione sostenuta: il paziente deve monitorare feedback visivi o uditivi forniti in tempo reale e adattare dinamicamente i propri parametri di cammino. Il BF stimola quindi funzioni

cognitive di tipo attentivo più che esecutivo. Studi che utilizzano BF durante il cammino su treadmill mostrano miglioramenti significativi nella lunghezza del passo, nella simmetria, nella regolarità del cammino e nella stabilità posturale (Carpinella et al., 2017; Ginis et al., 2016; Huang et al., 2006; McMaster et al., 2022). A differenza dei segnali ritmici open-loop, il BF opera in modalità closed-loop, consentendo correzioni motorie immediate e facilitando processi di apprendimento sensoriomotorio anche in presenza di deficit dei gangli della base (Muthukrishnan et al., 2019).

In questo quadro, un test di cammino in DT con livelli standardizzati e crescenti di difficoltà cognitiva può rappresentare uno strumento particolarmente utile per studiare in modo sistematico come il carico cognitivo influenzi il cammino e per valutare gli effetti di interventi che sfruttano compiti attentivi durante la marcia. Ciò è particolarmente rilevante per protocolli che richiedono un monitoraggio continuo e una modulazione volontaria dei parametri dell'andatura, come il cammino su treadmill con BF, nei quali il soggetto deve adattare passo, cadenza o simmetria in risposta a informazioni visive o uditive in tempo reale (Carpinella et al., 2017; McMaster et al., 2022). Studi che combinano treadmill training, compiti cognitivi o realtà aumentata hanno mostrato miglioramenti nelle prestazioni di cammino in DT, nella velocità e nella stabilità, oltre a una riduzione del rischio di caduta, confermando che il DT gait test può fungere sia da stimolo riabilitativo sia da outcome sensibile per il cambiamento (Kim et al., 2022; Mylius et al., 2021; Schaeffer et al., 2019; Vieira et al., 2014). Numerose evidenze mostrano inoltre che l'esercizio-terapia rappresenta un approccio efficace per migliorare il cammino e rallentare il deterioramento motorio nella MP, soprattutto quando l'allenamento è intensivo, personalizzato e avviato precocemente (Frazzitta et al., 2009; van der Kolk et al., 2019).

Nel loro complesso, queste evidenze suggeriscono che un test di cammino in DT con livelli graduati di difficoltà cognitiva possa costituire un banco di prova ideale per valutare l'efficacia di allenamenti di cammino su treadmill con BF, che integrano componenti aerobiche e attentive e mirano a migliorare sia l'automatizzazione locomotoria sia la capacità di gestire compiti cognitivi concorrenti. Tale approccio risulta coerente con la letteratura che sottolinea il ruolo dell'intensità dell'esercizio e della complessità del compito nel determinare adattamenti significativi a livello sia motorio sia cognitivo (MacInnis and Gibala, 2017; Zheng et al., 2021).

Obiettivo dello studio

L'obiettivo generale di questo lavoro è sviluppare strumenti di valutazione e interventi riabilitativi basati sull'evidenza per migliorare i parametri spazio-temporali dell'andatura nella malattia di Parkinson, con un focus specifico sul ruolo del DT e dell'allenamento su treadmill con BF. In particolare, gli obiettivi specifici dello studio sono i seguenti:

Sviluppare e validare un test standardizzato di cammino in DT con tre livelli crescenti di difficoltà cognitiva (sottrazioni in serie: -1, -3, -7). Tale test mira a quantificare in modo riproducibile e clinicamente rilevante l'effetto dell'aumento del carico cognitivo sui parametri spazio-temporali del cammino e a misurare i costi del DT nelle diverse condizioni, fornendo uno strumento sensibile per valutare l'automatizzazione dell'andatura e le sue modificazioni in risposta a interventi riabilitativi.

Valutare come il carico cognitivo influenzi il cammino nella MP in fase iniziale, confrontandolo con un gruppo di controlli sani coetanei. L'obiettivo è identificare differenze tra i gruppi nei parametri locomotori e nei costi DT, al fine di comprendere in che misura la perdita di automatizzazione locomotoria sia già presente nelle fasi precoci di malattia e se il cammino in DT possa essere utile per valutare compromissioni funzionali.

Verificare l'efficacia di un protocollo di allenamento aerobico su treadmill con BF nel migliorare i parametri di cammino e la fitness aerobica in persone con MP. Verranno analizzati gli effetti sulla velocità, sulla lunghezza del passo, sulla stabilità locomotoria, in condizioni di single- e DT; nonché l'eventuale relazione tra l'intensità dell'esercizio (misurata tramite il consumo di ossigeno, $\dot{V}O_2$) e i miglioramenti di fitness aerobica.

ABSTRACT

Parkinson's disease (PD) is a progressive neurodegenerative disorder characterised by degeneration of dopaminergic neurons in the substantia nigra, resulting in bradykinesia, rigidity, tremor, and marked gait disturbances. (Bloem et al., 2021; Tolosa et al., 2006). Spatiotemporal gait alterations, including reduced step length, increased cadence, decreased gait speed, and prolonged double support time, represent key factors contributing to impaired mobility and reduced functional independence in PD. (Abbruzzese et al., 2016; Tomlinson et al., 2014). In PD, gait automaticity is reduced due to impairments in motor learning and sensory feedback-based control, resulting in compensatory reliance on anticipatory (feedforward) mechanisms. (Nieuwboer et al., 2009; Olson et al., 2019). These deficits become especially apparent under dual-task (DT) conditions, where performing a cognitive task (such as counting or serial subtractions) while walking further disrupts gait spatiotemporal parameters (Kelly et al., 2012; Rochester et al., 2005). DT gait assessment, therefore, serves as a sensitive marker of impaired automatic walking. Recent research indicates that both single- and DT gait tests can identify early locomotor deficits in PD and distinguish patients from age-matched healthy controls, implying their utility in diagnosis and prognosis value (Caronni et al., 2025; Zhang et al., 2022).

In parallel, a growing body of evidence suggests that DT gait is also a valuable measure for examining the effects of targeted rehabilitation interventions. Importantly, two training modalities involving cognitive components operate through different mechanisms: DT training and biofeedback (BF) training. DT training combines gait with an additional cognitive task (e.g., counting, serial subtraction, verbal fluency). In this approach, the same cognitive functions assessed during DT testing, divided attention, working memory, and executive control, are directly targeted. Systematic reviews and randomised controlled trials demonstrate that this method can enhance DT performance, reduce DT costs, and improve gait speed, step length, and stability in individuals with PD (Beck et al., 2018; Johansson et al., 2022; Sarasso et al., 2024; Wollesen et al., 2021; Yang et al., 2019; Zheng et al., 2021).

BF training, by contrast, does not involve performing a separate cognitive task; instead, it requires a high level of sustained attention and concentration. Patients must continuously monitor visual or auditory feedback delivered in real-time and adapt their gait parameters accordingly. BF mainly stimulates attentional rather than executive cognitive processes. Studies using treadmill-based BF report significant improvements in step length, gait symmetry, gait regularity, and postural stability (Carpinella et al., 2017; Ginis et al., 2016; Huang et al., 2006; McMaster et al., 2022). Unlike open-loop rhythmic cueing, BF functions in a closed-loop mode, allowing for immediate motor adjustments

and supporting sensorimotor learning even when basal ganglia dysfunction is present (Muthukrishnan et al., 2019).

Within this framework, a DT gait test that includes standardised and progressively challenging cognitive conditions can be a particularly useful tool to systematically evaluate how cognitive load influences gait and to assess the impact of interventions that depend on attentional demands during walking. This is especially pertinent for protocols requiring continuous monitoring and voluntary adjustment of gait parameters, such as treadmill training with BF, where individuals must modify step length, cadence, or symmetry in response to real-time visual or auditory cues feedback (Carpinella et al., 2017; McMaster et al., 2022). Research involving treadmill training, cognitive tasks, Biofeedback or augmented reality has shown improvements in DT gait performance, gait speed, and stability, as well as a decrease in fall risk. This confirms that the DT gait test can serve both as a rehabilitative stimulus and a sensitive clinical outcome measure (Kim et al., 2022; Mylius et al., 2021; Schaeffer et al., 2019; Vieira et al., 2014). Further evidence suggests that exercise therapy is an effective method for enhancing gait and reducing motor decline in PD, especially when training is intensive, personalised, and started early (Frazzitta et al., 2009; van der Kolk et al., 2019).

Taken together, these findings indicate that a DT gait test with varying cognitive difficulty might be an ideal framework for assessing the effectiveness of treadmill-based BF gait training. This training combines aerobic and attentional elements and aims to improve both locomotor automaticity and the capacity to handle concurrent cognitive tasks. This approach aligns with the literature highlighting the importance of exercise intensity and task complexity in fostering meaningful motor and cognitive adaptations (MacInnis and Gibala, 2017; Zheng et al., 2021).

Aims of the study

The overall aim of this work is to develop evidence-based assessment tools and rehabilitation interventions to improve spatiotemporal gait parameters in PD, with a particular focus on the role of DT performance and treadmill training with BF. The specific objectives are:

Develop and validate a standardised ST gait test with three levels of increasing cognitive difficulty (serial subtractions: -1, -3, -7). This test aims to quantify in a reproducible and clinically meaningful way how increasing cognitive load affects gait spatiotemporal parameters and to measure DT costs across conditions, providing a sensitive tool to evaluate gait automaticity and its changes in response to rehabilitation.

Examine how cognitive load affects gait in early-stage PD compared with age-matched healthy controls, to identify between-group differences in locomotor parameters and DT costs, determining the extent to which reduced gait automaticity is already present in early disease, and assessing whether DT gait may serve as a sensitive marker of functional impairment.

To assess the effectiveness of an aerobic treadmill training protocol with BF in enhancing gait parameters and aerobic fitness in individuals with PD, this analysis will examine its effects on gait speed, step length, and locomotor stability under both single- and DT conditions. Additionally, it will explore the relationship between exercise intensity (measured via oxygen consumption, $\dot{V}O_2$) and improvements in aerobic fitness.

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CHAPTER 1: Dual-Task Gait Test Parkinson's Disease: Guidelines for Exercise Therapies and Treadmill with Biofeedback

1.1 Background of Parkinson's Disease

Parkinson's disease (PD) is a progressive neurodegenerative disorder caused by the degeneration of dopaminergic neurons in the substantia nigra. This neuronal loss results in a wide range of motor and non-motor symptoms that gradually impair functional independence and quality of life (Abbruzzese et al., 2016; Li et al., 2025; Tomlinson et al., 2014). The motor symptoms, including bradykinesia, rigidity, tremor, and gait impairment, reflect disruptions in basal ganglia circuits essential for the automatic regulation of voluntary movement (Abbruzzese et al., 2016; Frazzitta et al., 2009). As these primary motor issues advance, they are often worsened by additional symptoms such as postural instability, deficits in fine motor coordination, freezing of gait, dysphagia, and sleep disturbances, all of which further aggravate disability and increase the risk of falls (Bloem et al., 2021; Grimbergen et al., 2004; Tomlinson et al., 2014). Parallel to these motor complications, non-motor symptoms such as depression, cognitive decline, autonomic dysfunction, and speech or writing disturbances emerge, further adding to the overall clinical burden and often resulting in social withdrawal and diminished psychological well-being (Abbruzzese et al., 2016; Li et al., 2025). Understanding the neurophysiological mechanisms behind gait impairment offers vital insights into disease progression, showing how dopaminergic deficits interfere with basal ganglia-thalamus-cortical loops, affecting gait generation, scaling, and automatisation (Mazzoni et al., 2007). Consequently, individuals with PD rely heavily on cognitive and executive resources to control walking, which emphasises the increased effort needed for gait (Yogev-Seligmann et al., 2008). Neuroimaging studies have further revealed abnormal activation in brain regions such as the supplementary motor area, premotor cortex, and pedunculopontine nucleus (PPN) during gait tasks, indicating dysfunctional internal cueing mechanisms and impairments in gait initiation and rhythmic stepping. (Karachi et al., 2010). This dysfunction in the motor control network is worsened by structural and functional changes in the PPN, a key brainstem region involved in locomotion, which have been specifically linked to postural instability and freezing of gait (Thevathasan et al., 2018). Sensory integration abnormalities also contribute to gait decline; PD patients display altered proprioceptive weighting and a reduced ability to utilise sensory feedback for real-time gait adjustments, further increasing gait variability and reducing adaptability, particularly in challenging situations such as turning, obstacles, or multitasking (Wright et al., 2010; Zia et al., 2000). These deficits require increased activation of executive and attentional networks, including the dorsolateral prefrontal cortex and frontoparietal circuits, during walking as a form of compensatory recruitment (Amboni et al., 2013; Plotnik et al., 2009). However, under increased cognitive load, such as during dual-task (DT) walking, these compensatory mechanisms often become inadequate, leading to a noticeable decline in gait performance. Collectively, these neurophysiological impairments illustrate that gait dysfunction in PD results from a complex disruption of motor automaticity, sensory processing, attentional control, and executive-motor coupling.

The progression of PD is usually classified using the Hoehn and Yahr (H & Y) scale, which rates disease severity from 0 (no symptoms) to 5 (severe disability and reliance on a wheelchair) (Hoehn and Yahr, 1967). To gain a comprehensive understanding of both motor and non-motor features, the H & Y scale is often supplemented by the Movement Disorder Society-Unified PD Rating Scale (MDS-UPDRS), which assesses daily living activities, motor performance, and complications (Goetz et al., 2004). Recognising the heterogeneity of symptoms and their progression across disease stages highlights the importance of personalised therapeutic approaches. Pharmacological treatment, especially dopaminergic replacement with levodopa, along with surgical options like Deep Brain Stimulation (DBS), remains central to managing symptoms. Nonetheless, non-pharmacological strategies, including physiotherapy, structured exercise, speech therapy, nutritional guidance, and psychological support, play a vital role in maintaining independence and enhancing long-term outcomes (Tomlinson et al., 2014). Among these, rehabilitation and individualised exercise therapy are especially vital for addressing gait and balance problems, which are among the most disabling motor symptoms (Abbruzzese et al., 2016; Jones et al., 2022). These personalised exercise programmes aim to combat gait deterioration, usually characterised by reduced step length, increased cadence, and freezing episodes, while also improving mobility, reducing fall risk, and supporting both physical and emotional well-being (Frazzitta et al., 2009; Li et al., 2021; Olson et al., 2019). In recent years, increasing evidence suggests that structured physical activity can not only enhance functional capacity but may also delay disease progression. Although more research is needed to identify optimal training methods and intensities at various stages and for specific conditions, the potential benefits remain encouraging (Abbruzzese et al., 2016; Fernandes et al., 2020; Kathia et al., 2024).

1.2 Exercise Prescription Dose in Individuals Affected by Parkinson’s Disease

Exercise therapy is vital in managing PD, as it aims to maintain or increase physical activity, reduce stiffness and bradykinesia, and enhance fluidity of movement, gait coordination, and balance, ultimately supporting functional independence and residual mobility (Tomlinson et al., 2014). Evidence from multiple experimental and clinical studies supports the effectiveness of exercise in enhancing motor performance and quality of life for individuals with PD in the short and medium term (Abbruzzese et al., 2016; Fernandez del Olmo and Cudeiro, 2003; Frazzitta et al., 2009;

Nieuwboer et al., 2009). Contemporary recommendations emphasise a multidimensional approach that combines aerobic, resistance, flexibility, strength, balance, agility, multitasking, and cognitive training, all tailored to the patient's clinical profile and disease stage (Alberts and Rosenfeldt, 2020). According to the American College of Sports Medicine (ACSM, Liguori et al., 2022) and subsequent updates, exercise prescription for PD includes four core components. Aerobic activity is advised at least three days a week, with 30-minute sessions of continuous or intermittent moderate-to-vigorous intensity exercise such as walking, running, cycling, or swimming (60-80% heart rate reserve or 70-85% of maximum heart rate; or 14-17 on the 20-point RPE scale) (Alberts and Rosenfeldt, 2020; ACSM Liguori et al., 2022). Strength training should be performed two to three times a week, focusing on major muscles of the upper and lower limbs, with 10-15 repetitions per set using light to moderate resistance (ACSM Liguori et al., 2022). Balance, agility, and multitasking exercises are recommended two to three days per week, ideally with daily practice of multidirectional stepping, weight shifting, and dynamic balance activities, as well as exercises such as yoga, tai chi, dance, or boxing. Finally, stretching should be performed two to three days per week, with daily stretching offering the greatest benefit (ACSM Liguori et al., 2022).

Despite these clear guidelines, many healthy older adults do not meet the minimum activity levels, and physical activity declines even further after PD diagnosis (Cavanaugh et al., 2015). This underscores the importance of refining exercise prescription principles, specifically frequency, intensity, time, and type (FITT), to ensure evidence-based applications in PD. Limited understanding of optimal dosing often results in either the lack of a tailored exercise plan or in generic, non-individualised programmes that fail to deliver clinically meaningful improvements in cardiorespiratory fitness, balance, or cognition (Alushi et al., 2022). For this reason, recent research efforts aim to better characterise FITT principles in PD to enable more precise and effective exercise prescriptions.

Given the heterogeneity of symptoms and their progression, personalised and symptom-specific exercise programmes are vital. Tailored interventions consistently show better results compared to standard exercise protocols (Abbruzzese et al., 2016; Frazzitta et al., 2009; Zippenfening et al., 2023). According to Bloem (Bloem et al., 2001) individualised programmes should include structured initial and final assessments of physical and cognitive function (e.g., VO_2max, HRmax, MMSE), realistic patient-centred goals, targeted exercise selection (e.g., step length training, cadence regulation, balance improvement), and ongoing monitoring to ensure safety, effectiveness, and correct exercise intensity (e.g., VO_2 monitoring during training).

Several studies exemplify the benefits of individualised approaches. Frazzitta et al. (2009) demonstrated that a personalised treadmill gait training programme (20 minutes per day for 4 weeks,

with speed and step length progression every three days) resulted in greater improvements in gait speed, stride length, and 6MWT performance compared to non-tailored conventional approaches. Similarly, Feng et al. (2019) reported that personalised virtual reality balance training yielded greater improvements than traditional physiotherapy. Van der Kolk et al. (2019) found that individualised aerobic cycling, prescribed using a personalised heart rate reserve zone (50-80% HRR), significantly improved cardiorespiratory fitness, whereas a control physiotherapy programme did not.
These findings underscore the importance of structuring exercise therapy around individual clinical characteristics to maximise benefits. In this context, Alberts and Rosenfeldt (2020) proposed a theoretical model aimed at phenotyping patient responses to exercise based on demographic variables, exercise performance, and disease duration. This model categorises individuals as responders, neutrals, or non-responders based on changes in MDS-UPDRS III scores following aerobic cycling. The authors hypothesise that individuals with higher training heart rate percentages, higher cycling cadence, and shorter disease duration experience greater motor improvements. Such insights highlight the need for personalised, evidence-based prescriptions that integrate disease duration, fitness level, and training intensity to optimise motor and non-motor outcomes. Consequently, future research should focus on elucidating dose-response relationships in PD exercise therapy to guide clinical decision-making and rehabilitation strategies (Kathia et al., 2024; MacInnis and Gibala, 2017).

1.3 Direction of Current Research

Despite increasing evidence supporting exercise as a therapeutic approach in PD, various significant gaps hinder the development of fully evidence-based and personalised training protocols. Recent reviews highlight that the dose-response relationship of exercise, especially regarding optimal intensity thresholds, progression criteria, and long-term adherence, remains poorly understood (Panassollo et al., 2024). Furthermore, the neurobiological mechanisms by which various exercise modalities promote motor learning, neuroplasticity and gait improvements are still not fully understood, particularly in relation to disease stage and phenotype (Feng et al., 2020). Another major limitation is the heterogeneity of outcome measures and training protocols across studies, which hinders comparability and prevents the development of standardised guidelines (Yorke et al., 2021). Recent trials also emphasise the need for more rigorous stratification strategies that can identify patient subgroups who may respond differently to specific exercise modalities, intensities, or motor-cognitive interventions (Bouça-Machado et al., 2020; Wüllner et al., 2023).

Finally, the long-term sustainability of exercise-induced benefits, as well as the interaction between physical activity, disease progression, and non-motor symptoms such as cognitive decline and fatigue, remains insufficiently examined (Ng et al., 2025; Schenkman et al., 2018). Together, these gaps highlight the urgent need to develop mechanistically informed, personalised, and intensity-specific exercise prescriptions to optimise rehabilitation outcomes in PD.

These research gaps underline not only the need to refine exercise prescription in PD, but also the importance of developing sensitive and ecologically valid assessment tools capable of capturing motor-cognitive interactions during gait, which will be discussed in the following sections.

1.4 Gait Impairment in Parkinson's Disease: A Multidimensional Challenge

Gait disturbances are among the most disabling symptoms of PD, significantly impacting independence, quality of life, and risk of falls (Morris et al., 2001; Schaafsma et al., 2003). People with PD gait usually show a shorter stride, slower walking speed, more variability between strides, reduced arm swing, and changes in foot clearance (Hausdorff et al., 1998; Morris et al., 1996). These abnormalities lead to a high rate of falls, occurring in up to 60% of individuals with PD annually (Bloem et al., 2004; Grimbergen et al., 2004).

Crucially, gait impairment in PD does not stem from a single deficit but results from the interaction of several disrupted mechanisms. Basal ganglia dysfunction impairs internal cueing and the automatic selection of motor programmes (Morris et al., 1996), while impaired movement scaling decreases the ability to adjust step amplitude and adapt locomotion to environmental restrictions (Santens et al., 2003). Executive and attentional difficulties further reduce the ability to coordinate gait under cognitively demanding or unpredictable circumstances (Yogev-Seligmann et al., 2008). Altered sensorimotor integration, particularly the diminished ability to combine proprioceptive and visual cues, hampers real-time adjustments and leads to instability and inconsistent stepping (Azulay et al., 2002). Additionally, axial motor deficits and impaired postural responses compromise dynamic balance and heighten vulnerability during transitions, such as turning, starting, and stopping (Bloem et al., 2021).

Together, these factors create a complex gait disorder that challenges both assessment and intervention. Because gait performance in PD arises from the continuous interaction of motor, cognitive, and sensory systems, deficits can vary significantly between individuals and fluctuate in response to environmental and task demands. As a result, traditional single-task assessments, which evaluate walking in relatively simple and predictable contexts, may underestimate the true extent of gait impairment and its impact on everyday mobility.

This complexity highlights the need for assessment methods that can probe gait control under more realistic conditions, including situations that impose concurrent cognitive load and dynamic balance requirements. In this context, DT gait tests, which combine walking with a simultaneous cognitive task, have gained prominence as ecologically valid paradigms for quantifying motor-cognitive interference and reduced gait automaticity in PD. By comparing performance between single- and DT conditions, these protocols can reveal subtle deficits in step regulation, variability, and prioritisation strategies that are not apparent during simple walking.

Building on this reasoning, the rest of this chapter will discuss two closely related areas. The first relates to assessing gait performance during DT conditions. The second focuses on gait training through treadmill-based interventions with real-time biofeedback (BF). In the next chapter, these two areas, DT gait assessment and treadmill gait training with BF, will be explored in detail.

1.5 Introduction to Gait Testing

Gait testing is an essential tool for assessing mobility, fall risk, and motor performance in ageing and neurological conditions. In PD, quantitative gait analysis provides sensitive markers of impaired motor automaticity, reduced step regulation, and increased gait variability, features that are often overlooked during routine clinical assessments (Hausdorff et al., 1998). One of the earliest identified findings is that individuals with PD show increased stride-time variability compared to healthy adults, indicating disrupted locomotor rhythmicity and basal ganglia dysfunction (Hausdorff et al., 2001).

Standard clinical gait assessments, such as the 10-Metre Walk Test (10MWT) and the Timed Up and Go (TUG), reliably measure gait speed and functional mobility, and are widely utilised in rehabilitation and clinical trials (Podsiadlo and Richardson, 1991; Steffen and Seney, 2008). Gait speed, in particular, is regarded as a “functional vital sign” because of its ability to predict disability, falls, hospitalisation, and mortality among older adults (Del Din et al., 2016). These tests are straightforward to administer, reproducible, and sensitive to change.

Technological advancements have significantly improved the accuracy and ecological validity of gait assessment. Wearable inertial measurement units (IMUs) provide consistent spatiotemporal and kinematic gait parameters in both laboratory and real-world settings, enabling researchers to detect subtle PD-related impairments such as reduced arm swing and abnormalities in foot strike patterns (Del Din et al., 2016; Prisco et al., 2024). IMUs also facilitate extended monitoring in daily life, giving insights into real-world mobility that cannot be obtained through short clinic-based assessments (Rispens et al., 2015). In parallel, instrumented walkways and optoelectronic systems provide high-resolution measurements of step length, step time, variability, and spatial-temporal asymmetries over repeated strides in controlled environments. These systems provide precise characterisation of gait patterns and allow the validation of wearable-based measures (Lienhard et al., 2013; Webster et al., 2005).

Overall, gait testing, whether through traditional clinical tests or modern sensor-based approaches, provides an essential and objective method for evaluating mobility, tracking disease progression, and identifying early changes in motor control.

1.5.1 Limitations of Traditional Single-Task Gait Assessment

Despite the clinical importance of gait impairment, traditional assessments often rely on single-task walking, i.e., walking without any concurrent cognitive or motor demands. Although informative for evaluating basic gait performance, single-task assessments present several limitations that reduce their ecological and diagnostic relevance.

First, single-task gait lacks ecological validity: in everyday life, individuals rarely walk in isolation, but instead navigate busy environments, respond to sensory stimuli, converse, or carry objects, all of which place cognitive demands on walking (Yogev-Seligmann et al., 2008). Second, in early-stage PD, compensatory attentional strategies may enable patients to maintain near-normal single-task gait performance, creating plateau effects that mask subtle deficits (Rochester et al., 2004). Thirdly, single-task metrics underestimate fall risk, as falls in PD often occur in situations requiring divided attention, such as turning while talking or navigating through complex environments (Springer et al., 2006). Finally, traditional assessments neglect the critical role of cognition in gait control, overlooking executive dysfunction that significantly affects mobility in many individuals with PD (Amboni et al., 2013).

Together, these limitations emphasise the need for assessment frameworks that capture both motor and cognitive contributions to gait control.

1.5.2 Dual-Task Paradigm and Its Theoretical Rationale in Parkinson's Disease

The DT paradigm assesses the ability to perform two tasks simultaneously, typically walking while performing a cognitive task, and is based on cognitive theories that emphasise the limits of attentional capacity and resource allocation (Egeth and Kahneman, 1975; Wickens, 2002). When tasks contend for shared resources, performance declines, resulting in a measurable dual-task cost (DTC)

(Abernethy, 1988). DT paradigms are therefore especially suitable for evaluating motor-cognitive interactions and finding vulnerabilities that remain hidden under simple walking conditions.

In PD, the rationale for DT gait assessment arises from three converging factors. First, gait in PD is more attention-dependent than in healthy ageing: reduced automaticity increases reliance on executive and attentional resources, making walking particularly vulnerable to interference from concurrent cognitive activity (Morris et al., 1996; Wu and Hallett, 2008). Second, executive deficits, especially in set-shifting, inhibition, and working memory, are common even in early PD and impair the ability to coordinate simultaneous motor and cognitive tasks (Kudlicka et al., 2011; Lezak et al., 2012). Consequently, PD individuals generally show disproportionately high DT costs in both gait and cognitive performance compared with healthy controls (Yogev-Seligmann et al., 2008). Third, DT conditions better mirror real-world mobility demands than single-task walking. Everyday movement often involves simultaneous cognitive tasks, such as talking, planning routes, and crossing environmental obstacles, making DT paradigms more ecologically valid and predictive of instability, reduced adaptability, and fall risk (Beauchet et al., 2009; Springer et al., 2006). Early PD often appears normal during simple walking but shows clear deterioration under DT load, indicating that these paradigms reveal deficits masked by compensatory strategies during single-task conditions (Amboni et al., 2013).

Overall, the DT paradigm offers a sensitive window into motor-cognitive integration in PD. By challenging attentional and executive systems during gait, it detects subtle impairments, enhances diagnostic sensitivity, and underscores mechanisms relevant for prognosis and intervention planning.

1.5.3 Dual-Task Theoretical Frameworks

DT interference, the decline in performance when two tasks are executed concurrently, has been explained through several cognitive frameworks that help interpret DT gait deficits in PD. Capacity theories propose that individuals possess limited attentional resources; when concurrent tasks exceed available capacity, performance deteriorates (Egeth and Kahneman, 1975). In PD, reduced automaticity and increased attentional demands for gait exacerbate this limitation, resulting in disproportionately higher DT costs.

Multiple resource theory refines this view by suggesting that interference depends on how much two tasks draw on overlapping processing resources (Wickens, 2002). Because gait in PD requires increased cognitive support, even verbal tasks, normally minimally interfering, can compete with gait control, leading to measurable decrements.

Structural block models propose that certain cognitive operations, particularly response selection, must occur sequentially (Pashler, 1994; Welford, 1952). When gait adjustments and cognitive responses simultaneously require access to these central processes, delays emerge. Although direct evidence in PD is limited, this framework helps explain slowed reactions and increased variability during DT walking.

Executive control theories emphasise deficits in task coordination, inhibition, and set-shifting, domains frequently impaired in PD (Kudlicka et al., 2011). These difficulties limit the capacity to flexibly allocate resources between gait and cognitive tasks, contributing to the large and variable DT costs observed in PD.

Finally, compensatory neural mechanisms highlight the increased reliance on frontal cortical areas for gait in PD. Under DT load, these compensatory systems become overloaded, resulting in reduced gait stability and cognitive performance (Maidan et al., 2016; Wu and Hallett, 2008).

Together, these frameworks provide complementary perspectives: DT gait deficits in PD arise from reduced resources, inefficient allocation, structural processing delays, and impaired executive coordination, all of which are compounded by a limited compensatory capacity. This integrated view supports the use of DT paradigms to probe motor-cognitive interaction in PD.

1.5.4 Integrated Theoretical Model

DT interference has been described from various theoretical perspectives, including limited-capacity models, resource allocation frameworks, bottleneck theories, and executive control accounts. In PD, these mechanisms interact with disease-related disruptions in motor automaticity and executive functioning, leading to a notable vulnerability to interference during complex gait tasks (Egeth and Kahneman, 1975; Pashler, 1994; Welford, 1952; Wickens, 2002) .

From an integrated perspective, DT gait deficits in PD stem from four converging factors. First, diminished automaticity heightens the attentional demands of walking, leaving fewer resources available for concurrent cognitive tasks (Wu and Hallett, 2008). Second, impairments in inhibition, set-shifting, and task coordination, which are frequent even in non-demented PD, limit efficient DT performance management (Kudlicka et al., 2011; Lezak et al., 2012). Third, competition for central processing stages may delay both cognitive responses and gait adjustments, consistent with bottleneck models (Pashler, 1994). Finally, compensatory overactivation of prefrontal areas, already necessary during simple walking, becomes inadequate under DT load, leading to declines in both gait and cognitive performance (Maidan et al., 2016).

Although this integrated framework enhances understanding of DT interference in PD, significant gaps still exist. Most studies depend on indirect behavioural evidence rather than explicitly comparing competing theories, and the neural correlates of DT walking, especially the interaction between frontal, basal ganglia, cerebellar, and cholinergic systems, are only partly understood (Wu and Hallett, 2008). Individual variability is another challenge: patients with similar motor severity often show widely different DT costs (DTC), reflecting heterogeneity in cognitive profile, compensatory strategies, and sensory integration (Wu and Hallett, 2009; Yogev-Seligmann et al., 2008). Longitudinal studies are necessary to clarify the relationship between DT deficits and clinical progression, fall risk, and cognitive decline.

This integrated model underscores the value of DT gait paradigms as sensitive probes of motor-cognitive interaction in PD. Because DT performance reflects the combined efficiency of resource allocation, automatic motor control, and executive coordination, it reveals deficits that remain undetected in simple walking. These principles also guide methodological choices: the selection of cognitive tasks targeting executive control, the inclusion of gait parameters sensitive to reduced automaticity (e.g., variability, stride regulation, arm swing), and the need to assess both motor and cognitive performance derive directly from the mechanisms implicated in DT interference (Strouwen et al., 2016; Vitorio et al., 2021).

In the rigorous assessment of cognitive-motor interference, a fundamental methodological distinction must be maintained between task-inherent complexity, execution efficiency, and the resulting interference cost. The Index of Difficulty (ID), traditionally formulated within the framework of Fitts' Law, serves as a task-centric parameter that quantifies the spatial and informational demands of a motor action (Fitts, 1954; Slifkin et al., 2014). It is definedy the logarithmic relationship between movement amplitude (A) and target width (W), expressed as: $ID = log_2 (2A/W)$ thereby describing the environmental constraints rather than the subject's capacity. Conversely, the Rate of Correct Responses (RCR), or Throughput, functions as an output-centric metric of absolute efficiency. By integrating both speed and accuracy into a single success-per-unit-time value-often calculated as $RCR = (\frac{correct\ answers}{Total\ Time})$ it provides a robust measure of the information processing rate while effectively accounting for the speed-accuracy trade-off (Thorne, 2006; Heraud et al., 2018).

Finally, the DTC is characterised as a relative interference metric, representing the percentage decline in performance from single-task to dual-task conditions (Yogev-Seligmann et al., 2008; Al-Yahya et al., 2011). Calculated using the formula: $DTC\ (\%) = \frac{-(DT-ST)}{ST} \times 100$ it quantifies the "attentional tax" paid by the central nervous system during concurrent task execution. While the RCR defines the absolute productivity of the subject under load, the DTC isolates the specific vulnerability of the individual’s neural networks to resource competition, reflecting the integrity of attentional reserves and the degree of gait automaticity in both healthy ageing and neurodegenerative populations (Al-Yahya et al., 2011; Bayot et al., 2020).

In summary, theoretical integration supports DT gait assessment as an ecologically valid method to characterise motor-cognitive deficits in PD. This foundation motivates the methodological considerations discussed in the next section, which examines task selection, gait parameters, measurement technologies, and standardised procedures for high-quality DT assessment.
Dual-Task Cost (DTC): This is a measure of the relative change in performance when transitioning from a single-task (ST) to a dual-task (DT) condition. It is typically expressed as a percentage and calculated using the formula:

$$DTC\ (\%) = \frac{-(DT - ST)}{ST} \times 100$$

DTC represents the behavioural consequence of resource competition or "bottleneck" delays during multitasking. A higher DTC indicates greater interference and a higher reliance on attentional resources for gait (Liu et al., 2018).

1.6 Methodological Approaches to Dual-Task Gait Assessment

DT gait paradigms have become a crucial method for measuring the interaction between walking and higher-level cognitive processes. In DT tests, a normal walking task is paired with a simultaneous cognitive activity, usually mental tracking or verbal fluency, while performance is compared to single-task walking to determine the extra attentional resources needed for gait control (Yogev-Seligmann et al., 2008). Changes in walking speed, step variability or rhythm, and sometimes cognitive task performance, are seen as signs of cognitive-motor interference and less automatic gait. Meta-analysis reveals that these paradigms are sensitive to age, cognitive health, and neurological conditions, with more demanding cognitive tasks (e.g., serial subtraction) resulting in the largest declines in gait speed and stability (Al-Yahya et al., 2011). At the same time, several differences in methods have been noted regarding the type of cognitive task, outcome measures, and reporting strategies. This highlights the need for clearly defined protocols, standardised instructions, and explicit calculation of DT effects when designing gait assessments (Al-Yahya et al., 2011; Yogev-Seligmann et al., 2008). In this context, DT gait testing provides a structured yet adaptable approach for investigating the relationship between movement and cognition in PD, forming the foundation for the assessment method used in the following chapters.

1.6.1 Cognitive Task Selection

The choice of cognitive task is crucial in DT gait assessment because different tasks place varying demands on attention and executive control, areas often impaired in PD. Tasks frequently used in DT paradigms, such as phonemic fluency or serial subtraction, are straightforward to administer and consistently challenge executive functions relevant to motor-cognitive interference (Yogev-Seligmann et al., 2008). Their sensitivity has been confirmed in large-scale sensor-based studies where, for instance, alternating alphabet recitation produced notable DT effects on gait kinematics (Vitorio et al., 2021).

More demanding tasks, including auditory Stroop or working-memory paradigms, assess higher executive load and have demonstrated responsiveness to training effects (D'Cruz et al., 2020). Motor complexity also influences interference: turning or obstacle overstepping heightens the impact of cognitive load more than straight walking (Peterson et al., 2016; Stuart et al., 2019). Consequently, when selecting cognitive tasks, consideration should be given to both clinical practicality and the specific motor-cognitive domains that the assessment aims to target.

1.6.2 Gait Parameters and Measurement Platforms

DT gait assessment requires parameters that are sensitive to changes in automaticity and stability. Traditional spatiotemporal metrics, gait speed, stride length, cadence, and double support time, retain clinical utility, but kinematic features like arm swing amplitude and foot or heel strike angles have demonstrated superior discriminative value, particularly in early PD and under cognitive load (Vitório et al., 2021; Zhang et al., 2022). Measures of gait variability, particularly in stride time and stride length, are crucial markers of impaired rhythmicity and increased fall risk, and they correlate with executive dysfunction (Kelly et al., 2012).

Wearable IMUs offer a clinically practical platform for DT assessment. Their multi-sensor arrangements enable the simultaneous recording of lower-limb spatiotemporal metrics, upper-limb kinematics, turning performance, and trunk motion, facilitating the detection of sensitive markers such as arm swing and foot strike angle (Vitório et al., 2021). Instrumented treadmills and split-belt systems provide controlled conditions ideal for rehabilitation trials (Seuthe et al., 2020). While electronic walkways offer reliable spatiotemporal data (Lienhard et al., 2013; Webster et al., 2005). Motion capture remains the gold standard for accuracy; however, it is impractical for routine clinical use.

1.6.3 Dual-Task Cost, Protocol Design, and Methodological Quality

DTC measures the proportional decline in performance when transitioning from single- to DT conditions in both motor and cognitive domains. Interpretation must consider baseline performance and prioritisation strategies, as individuals may prefer gait or cognitive accuracy depending on the situation (Kelly et al., 2012). Reporting both motor and cognitive DTC is therefore vital for understanding motor-cognitive trade-offs.

High-quality DT protocols share several features. They include baseline single-task assessments, standardised cognitive tasks, explicit prioritisation instructions, and both straight and turning segments to capture dynamic balance demands. Order effects and fatigue should be controlled, and the medication state must be consistent, given its influence on gait performance (Smulders et al., 2016).

Methodological consistency is essential. Studies should report all gait and cognitive parameters, describe the technology employed, and provide reference or normative data to contextualise performance (Strouwen et al., 2016). Reliability metrics, especially test-retest and inter-rater measures, are essential to ensure stability, as DT outcomes are influenced by attentional fluctuations and fatigue (Lord et al., 2014). While IMUs, walkways, treadmills, and motion capture systems all provide distinct advantages, validation studies comparing these platforms remain scarce, highlighting the necessity for standardised protocols and shared analytical frameworks (Mirelman et al., 2016a). Clear reporting of DTC formulas and protocol details remains an essential yet often neglected requirement (Plummer et al., 2015).

1.7 Clinical and Diagnostic Applications of Dual-Task Gait Assessment

DT gait assessment offers diagnostic value in PD because it detects impairments that appear when cognitive and motor demands need to be coordinated, conditions that closely resemble everyday mobility (Amboni et al., 2013; Yogev-Seligmann et al., 2008). During simple walking, early PD may seem relatively preserved due to compensatory strategies, whereas DT conditions reveal limitations in attentional allocation, gait regulation, and executive control.

1.7.1 Early Diagnostic Value and Disease Severity

DT paradigms improve early diagnostic sensitivity by challenging motor-cognitive integration. Even in single-task conditions, high-resolution kinematic markers such as stride length and heel strike angle can distinguish early PD from healthy ageing (Zhang et al., 2022). DT conditions accentuate these differences: parameters reflecting automatic gait control, such as arm swing amplitude, foot strike angle, gait variability, and turning performance, exhibit disproportionately large DT costs in PD and have strong discriminative accuracy (Stegemöller et al., 2014; Strouwen et al., 2016; Wu and Hallett, 2009).

DT impairment also worsens with disease severity. Greater DT costs in gait speed, stride length, variability, and especially turning measures are linked to Hoehn and Yahr stage and UPDRS-III motor severity (Curtze et al., 2015; Kelly et al., 2012a; Maidan et al., 2016). Although longitudinal evidence still remains limited, these findings endorse the use of DT gait metrics as additional indicators of disease progression and axial motor decline.

1.7.2 Fall Risk and Cognitive Profiling

Since most falls in PD happen under conditions that need divided attention or quick motor responses, DT gait assessment is especially useful for assessing fall risk (Bloem et al., 2004; Grimbergen et al., 2004). Increased gait variability, decreased stride length, and exaggerated DT costs have been linked to fall history and future fall incidence (Amboni et al., 2013; Heinzel et al., 2016; Kelly et al., 2012; Springer et al., 2006). DT turning impairments indicate vulnerability in dynamic balance and visuomotor coordination (Curtze et al., 2015; Maidan et al., 2016).

DT paradigms also offer a functional insight into cognitive status. Individuals with PD and mild cognitive impairment show greater motor and cognitive DT costs than their cognitively normal counterparts, even when matched for motor severity (Amboni et al., 2013; Nieuwhof et al., 2017; Strouwen et al., 2016). These findings suggest that clinically significant DT interference could indicate early executive dysfunction and emphasise the need for more thorough cognitive assessment.

1.7.3 Parkinson's Disease vs. Healthy Controls

Comparisons between individuals with PD and healthy older adults reveal distinct patterns of DT gait impairment that go beyond normal ageing. Quantitative gait analyses consistently show reduced stride length, slower walking velocity, diminished arm swing, increased variability, and altered foot strike patterns in PD, even at early stages of the disease (Hausdorff et al., 1998; Mirelman et al., 2016a; Morris et al., 1996). These abnormalities indicate a higher dependence on cognitive control for gait regulation in PD, rendering motor performance especially susceptible when attention is divided.

Under DT conditions, individuals with PD show disproportionately large reductions in gait speed, stride length, turning velocity, and stability compared with healthy controls (Kelly et al., 2012;

Yogev-Seligmann et al., 2008). Kinematic features such as arm swing amplitude and foot strike angle exhibit particularly strong DT sensitivity and discriminative power (Vitório et al., 2021). These differences reflect neurophysiological findings showing reduced automaticity and increased executive load during gait in PD.

Qualitative differences in task coordination also arise. Healthy older adults usually employ adaptive strategies, such as reducing cognitive load or altering step patterns, to maintain stability during DT walking. In contrast, individuals with PD often show inconsistent or inflexible prioritisation, sometimes favouring the cognitive aspect task ("cognitive priority") and at other times the motor task ("posture first"), reflecting impaired executive control and reduced flexibility in resource allocation (Bloem et al., 2004; Kudlicka et al., 2011). Disease severity and cognitive status further modulate DT performance. Early-stage PD already shows detectable DT costs, particularly in parameters that depend on automatic motor control (Zhang et al., 2022). As severity progresses, DT deficits worsen across spatiotemporal and kinematic domains, potentially preventing safe performance entirely (Mirelman et al., 2016a). Individuals with PD and mild cognitive impairment show even higher DT costs and increased turning instability compared to cognitively normal PD, emphasising the role of executive dysfunction in motor-cognitive interference (Amboni et al., 2013; Maidan et al., 2016).

Overall, DT gait assessment offers strong discriminative ability for differentiating PD from healthy ageing, especially useful in detecting early impairments and capturing the combined effects of reduced automaticity and executive dysfunction. Multi-parameter models that include stride length, gait variability, arm swing, foot strike angle, and turning metrics attain excellent classification results (Vitorio et al., 2021; Zhang et al., 2022).

1.7.4 Clinical Integration and Practical Considerations

For the DT gait assessment to be clinically meaningful, protocols must be standardised with respect to cognitive tasks, prioritisation instructions, walking distance, and inclusion of turning segments (Kelly et al., 2012a; Strouwen et al., 2016). Multi-sensor IMU systems are preferred because of their ability to accurately measure sensitive kinematic parameters with real-world relevance (Prisco et al., 2024). Interpretation should be based on normative values where available (Olivier Beauchet et al., 2009) and include additional clinical data, such as neurological examination, motor scales, and cognitive testing.

Medication status, fatigue, mood, and comorbidities influence performance and should be recorded or controlled (Smulders et al., 2016). Because DT metrics reflect integrated motor-cognitive functioning, repeated and standardised assessments enhance reliability and support their use in identifying early impairment, monitoring disease progression, and guiding targeted intervention strategies.

The effective clinical implementation of DT gait assessment in PD relies on adopting standardised protocols, validated measurement technologies, and clear interpretative criteria that guide the translation of experimental evidence into practical clinical use. Since DT performance reflects the interaction between gait automaticity, executive control, and attentional allocation, clinical assessments must be designed to reliably capture these interconnected components (Kelly et al., 2012a; Mirelman et al., 2016a). A coherent methodological approach guarantees that DT outcomes can serve diagnostic, prognostic, and rehabilitative functions.

A standardised DT protocol should combine both single-task and DT walking, including turning segments which impose greater cognitive-motor demands and are ecologically relevant to daily mobility. The cognitive tasks chosen, typically verbal fluency or serial subtraction, should reliably challenge executive and attentional functions known to be impaired in PD (Yogev-Seligmann et al., 2008). Clear prioritisation instructions are crucial, as motor-cognitive trade-offs significantly influence DT performance; without explicit guidance, interpreting DT costs becomes unreliable. Additionally, the length of the walking path is important: distances around 10-20 metres facilitate obtaining stable measures of gait variability and DT cost, reducing variability and enhancing reproducibility (Olivier Beauchet et al., 2009). Since gait automaticity and attentional demands fluctuate with dopaminergic levels, assessments should be conducted in a standardised medication state, ideally during the ON phase, although OFF-state testing can be useful for specific clinical questions (Smulders et al., 2016).

Environmental conditions must also be carefully managed. Variations in lighting, background noise, cues, or available space can cause significant inconsistency, especially in individuals with impaired attentional regulation. Tracking fatigue is equally important, as a decline in attention disproportionately impacts DT performance and can lead to misinterpretation, particularly in older or more severely affected patients.

From a technological perspective, IMUs currently provide the best balance between ecological validity and measurement resolution. Multi-sensor IMU configurations can capture detailed gait features that are highly sensitive to PD-related deficits, such as stride length, gait variability, arm swing amplitude, trunk motion, and turning metrics, while enabling assessment in real-world environments (Del Din et al., 2016; Prisco et al., 2024; Youssef et al., 2025). Although laboratory-based optical motion capture remains the gold standard for precise kinematics, its costs, operational demands, and limited ecological validity restrict its use in routine clinical practice. Pressure-sensitive walkways offer accessible and reliable spatiotemporal measurements, but cannot assess upper-limb kinematics, a parameter set that has demonstrated strong discriminative ability for identifying subtle gait abnormalities in PD (Youssef et al., 2025).

The interpretation of DT outcomes should be based on normative reference values whenever they are available. Age- and sex-specific norms are especially important for parameters such as gait variability and DT cost, which naturally increase with age even in healthy individuals (Olivier Beauchet et al., 2009). Comparing data against normative standards enhances the accuracy in detecting pathological patterns, clarifies whether performance indicates impaired motor-cognitive coordination or compensatory strategies, and improves the clinical usefulness of DT assessments for differential diagnosis and rehabilitation planning.

In summary, the clinical use of DT gait assessment requires standardised protocols, ecologically valid measurement systems, and reference-based interpretative frameworks. When applied rigorously, DT assessment provides a sensitive and functionally relevant insight into motor-cognitive integration in PD, complementing traditional clinical measures and aiding personalised clinical decision-making.

1.7.5 Treatment Outcomes and Intervention Responsiveness

DT gait metrics are highly sensitive markers of treatment response in PD, as they measure changes in motor automaticity, attentional coordination, and executive function that often go unnoticed in single-task conditions (Abbruzzese et al., 2016; Kelly et al., 2012). Because DT paradigms directly evaluate the ability to control gait while managing cognitive tasks simultaneously, improvements in DT performance serve as a practical indicator of better motor-cognitive integration and demonstrate how rehabilitation produces its effects.

Increasing evidence suggests that DT gait performance can be improved through interventions targeting gait automaticity, motor adaptation, or attentional control. For example, a single session of split-belt treadmill adaptation leads to immediate enhancements in spatiotemporal and kinematic aspects of DT walking, along with better concurrent Stroop performance, implying more efficient use of attentional resources and improved interlimb coordination (D’Cruz et al., 2020). Long-term exercise-based interventions also provide significant benefits. Intensive treadmill training with cueing (Frazzitta et al., 2009) increases stride amplitude and gait symmetry while decreasing reliance on compensatory attentional strategies, whereas high-intensity aerobic exercise boosts gait automaticity and cardiorespiratory fitness, indirectly supporting DT performance through improved physiological reserve (van der Kolk et al., 2019).

Beyond aerobic or treadmill-based interventions, motor-cognitive training paradigms, including attentional focus tasks, cueing strategies, and exercises requiring continuous adjustment of step timing or direction, demonstrate positive effects on gait variability during DT walking (Ginis et al., 2016; Olson et al., 2019). These improvements likely reflect strengthened executive processes involved in step regulation and more efficient integration between cognitive control and locomotor systems. Interventions that challenge dynamic adaptation, such as turning practice or variable-speed walking, appear particularly effective, probably because they place higher demands on motor planning and cognitive flexibility, both critical determinants of DT gait. Across studies, consistent patterns of treatment responsiveness emerge. Improvements typically include increased DT gait speed and stride length, more stable kinematic control, evident in enhanced arm swing and more consistent foot strike patterns, reduced stride-to-stride variability, and smaller motor DTC. Some interventions further improve cognitive-task accuracy during walking, indicating a more efficient sharing of resources between motor and cognitive domains rather than simply prioritising one domain over the other.

Importantly, these treatment effects are influenced by clinical characteristics. Patients in the early stages tend to exhibit greater and quicker improvements, probably due to preserved neural plasticity and less ingrained motor impairment (Frazzitta et al., 2009). Similarly, individuals with milder executive dysfunction respond more strongly to motor-cognitive training, whereas those with more significant cognitive impairment often show limited improvements in DT gait (Olson et al., 2019). Intervention intensity and task specificity are also key factors affecting outcomes: high-intensity and DT-specific programmes consistently lead to larger improvements than low-intensity or non-specific approaches (van der Kolk et al., 2019).

Taken together, these findings highlight the importance of a personalised rehabilitation framework where DT profiles, whether mainly motor, mainly cognitive, or mixed, guide both intervention choices and progression. By tailoring therapeutic strategies to individual patterns of DT impairment, clinicians can more effectively address the mechanisms behind motor-cognitive dysregulation in PD and improve functional mobility in everyday life.

1.8 Research Gaps and Future Directions for the Gait Dual-Task Test

Despite notable advances in understanding DT gait impairment in PD, significant gaps remain that restrict its full clinical application. A key challenge involves methodological heterogeneity, as studies vary significantly in cognitive task selection, walking distance, prioritisation instructions, and reported gait parameters, which complicates cross-study comparisons and hinders the development of standardised protocols (Kelly et al., 2012a; Mirelman et al., 2016a). Reporting practices are equally inconsistent, with motor and cognitive outcomes not always presented together and formulas for DT cost calculation sometimes omitted, which reduces transparency and interpretability (Plummer et al., 2015). Variability in measurement technologies, including IMUs, electronic walkways, treadmills, and motion-capture systems, further complicates synthesis and highlights the importance of validation studies that directly compare these platforms (Mirelman et al., 2016a). The absence of extensive normative datasets divided by age, sex, disease stage, and cognitive status remains a significant limitation, as such reference data are essential for meaningful clinical interpretation (Beauchet et al., 2009a; Strouwen et al., 2016).

Mechanistic understanding also remains incomplete. While behavioural studies consistently demonstrate exaggerated DTC in PD, the underlying neural pathways are only partially characterised. Interactions among basal ganglia circuits, fronto-striatal and cerebellar networks, and cholinergic systems during DT walking remain poorly understood, largely due to the technical challenges of capturing brain activity during ecological gait (Shine et al., 2013; Wu and Hallett, 2005). Emerging methods, such as fNIRS, mobile EEG, and multimodal approaches that combine neural and kinematic data, provide promising opportunities for studying motor-cognitive integration in more naturalistic settings (Clark et al., 2020; Maidan et al., 2016). Further research is necessary to clarify compensatory cortical mechanisms and to understand non-dopaminergic contributions, such as cholinergic degeneration, which has been linked to gait variability (Bohnen and Albin, 2011).

A further limitation lies in the scarcity of longitudinal evidence. Most DT gait studies employ cross-sectional designs, which limit our understanding of how DT impairments develop over time or predict future outcomes, such as cognitive decline and increased fall risk. Defining minimal clinically important differences and thresholds for clinically meaningful change in DT metrics is crucial for accurately interpreting the effects of interventions. Further research is necessary to determine whether improvements in DT performance translate to real-world mobility and if specific rehabilitation strategies, such as high-intensity, motor, or combined approaches, provide greater benefits (D'Cruz et al., 2020; van der Kolk et al., 2019). The increasing use of technologies for continuous monitoring may help close these gaps by enabling the assessment of DT performance in daily settings and supporting the development of personalised, adaptive rehabilitation strategies.

Overall, addressing these methodological, mechanistic and longitudinal gaps is vital for enhancing the diagnostic and prognostic utility of DT gait assessment in PD and for integrating it into evidence-based, personalised clinical pathways.

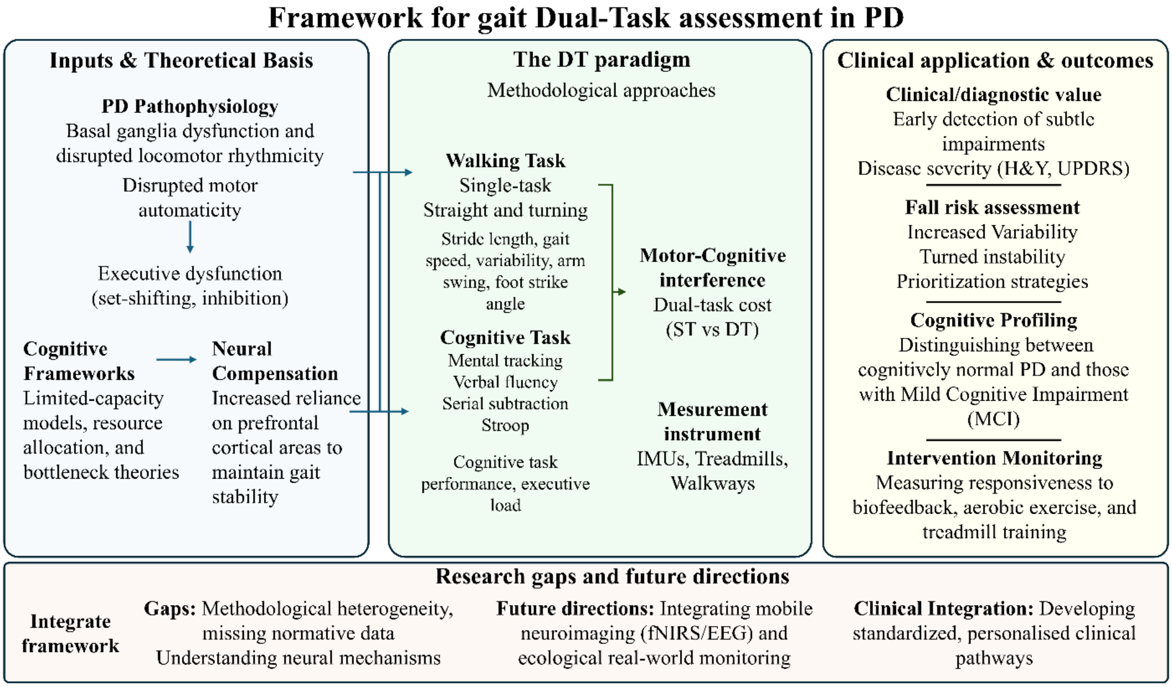
DT gait assessment provides a sensitive and ecologically relevant insight into the interaction between motor and cognitive systems in PD. Evidence demonstrates that increasing cognitive load disproportionately impacts spatiotemporal and kinematic gait parameters in PD, especially stride length, gait variability, arm swing, and foot strike angles, uncovering impairments that may remain hidden during simple walking conditions (Kelly et al., 2012a; Vitorio et al., 2021; Zhang et al., 2022). These findings illustrate the combined impact of disrupted automaticity, executive dysfunction, and compensatory prefrontal overactivation, supporting DT measures as valuable indicators of motor-cognitive integration.

A significant challenge highlighted by current evidence is the absence of a standardised DT gait protocol, despite its clear clinical importance in assessing gait under increasing cognitive load. The variation in DT tasks, walking distances, and measurement methods across studies hampers comparability, diminishes diagnostic accuracy, and hinders the development of reference values that could facilitate routine clinical application. There is an urgent need to create and validate a standardised DT protocol incorporating varying levels of cognitive difficulty to systematically evaluate impairments in gait automaticity and motor-cognitive integration, and to distinguish PD-related deficits from those found in healthy individuals ageing (Carpinella et al., 2017; van der Kolk et al., 2019).

Clinically, DT gait metrics show value for early detection, disease staging, fall-risk assessment, and personalising rehabilitation strategies. They complement neurological examination and traditional motor scales by providing objective measurement of how individuals handle cognitive load during walking, a key aspect of real-world mobility. Technology further enhances their usefulness by allowing high-resolution assessment in both laboratory and real-life environments.

Despite these strengths, challenges persist. Methodological variability has limited the longitudinal evidence, and an incomplete understanding of the neural mechanisms underlying DT impairment constrains its clinical application. Future research should focus on standardised protocols, multimodal neuroimaging techniques, and longer-term studies to identify clinically meaningful changes and assess their real-world relevance.

Overall, addressing these methodological, mechanistic, and longitudinal gaps is vital for enhancing the diagnostic and prognostic utility of DT gait assessment in PD and for integrating it into evidence-based, personalised clinical pathways. At the same time, the evidence reviewed in this section supports DT gait assessment as a robust and clinically relevant method for characterising motor-cognitive deficits in PD. These insights provide the conceptual and methodological foundation for the second part of this chapter, which will examine treadmill-based gait training interventions with real-time BF designed to target the specific impairments revealed by DT paradigms and to improve functional gait performance and independence.



1.9 Historical Evolution of Evidence on Treadmill Training with Biofeedback

Over the past three decades, research on gait rehabilitation in PD has shifted progressively from simple cueing approaches to more structured, technology-enhanced interventions that integrate sensory, cognitive, and motor information during walking. Early studies demonstrated that individuals with PD could respond meaningfully to external cues to modulate gait rhythm and amplitude, laying the foundation for subsequent developments in treadmill-based rehabilitation enriched with real-time feedback (Johnston, 2009; Schlick et al., 2012); as the field evolved, technological advances enabled the transition from static, open-loop cueing to closed-loop systems that deliver real-time visual or multimodal information during treadmill walking (Jellish et al., 2015; Studer et al., 2017). More recently, immersive and semi-immersive virtual reality (VR) environments have been introduced, enabling motor-cognitive treadmill training capable of targeting gait automaticity, balance, and fall risk more comprehensively (Mirelman et al., 2013; Mirelman et al., 2016b; Pelosin et al., 2022; Pullia et al., 2023).

The earliest uses of external cues in PD gait rehabilitation emerged in the 1990s and early 2000s, demonstrating that rhythmic auditory stimulation, visual stepping markers, and attentional strategies could help initiate gait and increase step size by compensating for impaired internal cueing mechanisms to some extent (Morris, 2010; Nieuwboer et al., 2007). These early findings supported the idea that sensory enhancement could temporarily make up for deficits in automatic motor control.

The shift towards treadmill-based interventions occurred when researchers began investigating the benefits of combining continuous walking with simple visual feedback or optical flow adjustments. These initial studies showed immediate improvements in gait symmetry, trunk alignment, and stride length, suggesting that treadmill settings offer a controlled, high-repetition platform suitable for feedback-based motor learning (Johnston, 2009; Schlick et al., 2012). Around the same time, advancements in motion-tracking technology enabled the first real-time dynamic visual cueing and kinematic displays, marking the beginning of closed-loop feedback systems for PD gait training (Ebersbach et al., 2010). These systems represented a shift from fixed cueing to interactive feedback, which can monitor performance and deliver targeted corrections. From the 2010s, a major expansion in clinical research introduced sensorised treadmills, wearable IMUs, and semi-immersive VR platforms (Jellish et al., 2015; Rudinskiy et al., 2014). Instrumented treadmills enabled precise monitoring of step length, cadence, foot trajectory, and force, allowing personalised feedback. VR-enhanced treadmill systems marked a second key phase by integrating crossing obstacles, visuospatial challenges, and DT elements, extending training beyond motor practice to engage cognitive and sensory (Mirelman et al., 2013; Pullia et al., 2023). This approach effectively addressed gait automaticity deficits and fall risk, with multicenter trials showing significant reductions in falls among PD patients (Tarsy et al., 2016). Recent advances include home-friendly, wearable systems that provide visual or auditory feedback via sensors, tablets, or projection, aiming to improve access and long-term practice; however, further research on clinical efficacy is needed (Luis-Martínez et al., 2020; Pelosin et al., 2022).

1.10 Theoretical Rationale for Treadmill Training with Biofeedback in Parkinson’s Disease

In PD, gait disturbances such as reduced stride length, increased step-to-step variability, diminished propulsive force, and compromised postural control stem from the progressive loss of automatic motor regulation. This loss of automaticity compels individuals to rely disproportionately on cognitive resources to maintain and adjust basic locomotor functions, especially in complex or unpredictable environments. Evidence accumulated over recent decades suggests that these deficits can be partially mitigated through structured exercise programmes that combine aerobic stimulation, task-specific practice, and external sensory information, which can compensate for or enhance internally impaired motor signals (Johnston, 2009; Mak et al., 2017; Schlick et al., 2012).

Within this framework, treadmill gait training combined with real-time BF has emerged as a promising intervention aimed at strengthening motor learning processes and improving gait performance in PD. The treadmill provides a highly repeatable environment that stabilises spatiotemporal parameters and facilitates consistent motor practice. BF adds a layer of dynamic, real-time sensory augmentation, heightening awareness of movement quality and encouraging continuous self-correction (Jellish et al., 2015; Studer et al., 2017). This combination closely aligns with the conceptual foundations of cueing-based therapies and cognitive feedback strategies, which have been shown to assist individuals with PD in compensating for deficits in internal rhythm generation and motor scaling (Pullia et al., 2023; Cornelia Schlick et al., 2012).

1.10.1 External Cueing and Real-Time Feedback as Supports for Impaired Automaticity

A fundamental reason for introducing BF during gait training stems from the longstanding evidence that external cues, whether visual, auditory, or multimodal, can compensate for impaired internal motor guidance. Traditional open-loop cues, such as rhythmic auditory stimulation or fixed visual markers, provide structured periodic information that helps regulate step timing or amplitude. However, closed-loop cues, which characterise BF systems, offer a distinctly different form of support: they detect performance in real-time and supply immediate sensory information that enables the user to monitor and adjust their performance continuously (Jellish et al., 2015).

This mechanism is similar to cognitive feedback approaches in PD rehabilitation, where increased attentional engagement and error monitoring are used to activate residual learning processes. Studies using real-time visual feedback during treadmill walking have shown immediate increases in step length, better trunk alignment, and reduced gait asymmetry, demonstrating that individuals with PD can effectively use externally provided sensory information to improve gait performance (Pullia et al., 2023; Schlick et al., 2012).

These findings support the broader hypothesis that external sensory inputs may bypass dysfunctional basal ganglia circuits by recruiting alternative neural pathways, particularly cortico-cerebellar networks, which remain capable of supporting motor regulation despite dopaminergic degeneration. This compensatory mechanism aligns with models suggesting that attentional strategies and sensory augmentation can temporarily replace impaired automatic control systems (Morris, 2010; Nieuwboer et al., 2007).

1.10.2 Treadmill Training as a High-Repetition Platform for Motor Learning

Treadmill-based exercise provides an ideal platform for motor learning, allowing for a high number of controlled, repeatable gait cycles. This repetitive pattern acts as a strong stimulus for neural adaptation, especially when combined with explicit feedback. The treadmill stabilises temporal fluctuations and synchronises rhythmicity, creating an environment where corrective cues can be more easily perceived, integrated, and reinforced over time (Byl, 2015; Gulcan et al., 2022).

When combined with BF, treadmill walking becomes a two-way interaction between motor output and sensory feedback. Through continual exposure to gait targets, such as foot placement, trunk angle, or step length, the participant engages in a form of guided motor practice that enhances the cognitive stage of learning and gradually encourages the automation of corrected movement strategies. This aligns with modern models of motor learning, indicating that explicit feedback accelerates initial learning, while repeated exposure during rhythmic practice facilitates the transition to automatic performance. The inclusion of cognitive elements within treadmill training, such as VR-based obstacle negotiation or visuospatial tasks, has been associated with broader improvements, including enhanced DT performance and a reduction in falls. (Mirelman et al., 2016a; Pelosin et al., 2022). These benefits suggest that treadmill + BF training not only improves the biomechanics of gait but also strengthens motor-cognitive interactions that are essential for safe mobility in daily life.

1.10.3 Sensorimotor Recalibration Through Augmented Feedback

PD is characterised by reduced proprioceptive accuracy and impaired integration of sensory information relevant to postural control and locomotion. BF systems address this deficit by providing highly relevant visual, auditory, or multimodal cues that can recalibrate internal representations of movement. Real-time visual feedback, such as projected foot trajectories, step targets, or kinematic avatars, enhances the salience of movement amplitude and timing, helping participants regain awareness of gait parameters that would otherwise be underestimated (Jellish et al., 2015; Rudinskiy et al., 2014).

VR enhanced treadmill systems further extend this approach by immersing the user in ecologically relevant contexts that involve crossing obstacles, pathways, and attentional challenges. Multicenter trials have demonstrated that VR-based treadmill training produces clinically meaningful improvements in fall risk and functional mobility beyond those achieved with treadmill training alone (Mirelman et al., 2013). These results support the hypothesis that enriched multisensory environments promote deeper sensorimotor integration, contributing to the consolidation of more stable and adaptive gait patterns.

The ability of visual and VR-based cues to improve propulsive force, step length, and postural adjustments may be linked to the enhanced engagement of cortical networks involved in visuomotor integration and attention (Keus et al., 2009; Konczak et al., 2009).

1.10.4 Cognitive Engagement and Motor - Cognitive Interaction During Gait

Because gait becomes progressively less automatic in PD, individuals must allocate a greater proportion of their attentional resources to maintain and regulate the locomotor rhythm. This increased cognitive load contributes to gait slowing, increased variability, and reduced adaptability, especially when performing additional cognitive tasks. The integration of BF into treadmill training reinforces attentional engagement, stimulates executive control mechanisms, and supports the error-correction processes that underlie motor learning (Pelosin et al., 2022).

VR-based treadmill paradigms that integrate sensory feedback with cognitive challenges, such as obstacle anticipation or visuospatial discrimination, have demonstrated potential in enhancing DT performance. These findings highlight the promise of motor-cognitive treadmill training to improve not only gait mechanics but also the cognitive abilities that support safe mobility in real-world settings (Mirelman et al., 2013; Mirelman et al., 2016b; Pullia et al., 2023).

This synergistic effect reflects the conceptual foundation of cognitive feedback therapy, where increased attentional demand and motivational engagement are hypothesised to activate preserved neural resources, promote the formation of alternative motor pathways, and ultimately improve the acquisition and retention of new motor skills(Abbruzzese et al., 2016; Carpinella et al., 2017).

1.10.5 Integrating Aerobic Stimulus and Biofeedback: A Multidimensional Approach

Treadmill training naturally offers an aerobic stimulus, which has been demonstrated to enhance gait endurance, cardiovascular health, and overall mobility in PD populations (Byl, 2015). When combined with BF, the treadmill becomes a multifaceted therapeutic instrument that simultaneously targets aerobic fitness, motor learning, sensorimotor integration, and cognitive control. This holistic approach reflects a growing perspective in PD rehabilitation: that successful interventions must consider the interaction between motor, cognitive, and sensory systems to combat the progressive decline of automatism (Pelosin et al., 2022; Pullia et al., 2023).

Overall, the theoretical rationale for treadmill training with BF suggests that this combined approach supports residual motor learning mechanisms, compensates for internal cueing deficits, increases awareness of gait parameters, and enhances motor-cognitive interactions. These mechanisms collectively contribute to improvements in gait stability, functional mobility, and reduction of fall risk, providing a comprehensive therapeutic strategy that aligns with the multifactorial nature of gait control impairments in PD.

1.11 Typologies of Biofeedback and Their Physiological Targets

In recent years, treadmill training supplemented with real-time BF has become an increasingly popular method to support gait rehabilitation in PD. This approach combines aerobic exercise with sensory augmentation systems designed to improve awareness of specific gait parameters and enhance motor learning processes that are impaired by basal ganglia dysfunction (Johnston, 2009; Schlick et al., 2012). Different types of BF focus on various physiological aspects of gait, such as step amplitude, rhythm regulation, trunk alignment, and propulsive force. Although these systems differ in complexity, from simple visual displays to immersive VR, they all aim to provide external information that can replace or reinforce internal motor cues that are compromised (Jellish et al., 2015; Studer et al., 2017). Below, the main categories of BF used in treadmill-based rehabilitation for PD are summarised alongside their primary physiological targets.

1.11.1 Visual Biofeedback and Modulation of Spatial Gait Parameters

Visual feedback is the most extensively studied modality and is widely used to address spatial deficits characteristic of PD gait, such as reduced step and stride length, gait asymmetry, and insufficient foot clearance. Real-time visual cues can be delivered through screen-based kinematic displays, projected step targets, or semi-immersive VR environments capable of manipulating optic flow and foot trajectory (Nagai et al., 2020; Rudinskiy et al., 2014).

Visual feedback has shown particular effectiveness in increasing step length and improving foot placement accuracy, likely because it enhances visuospatial awareness and compensates for proprioceptive deficits often seen in PD. By providing immediate feedback on step amplitude or trajectory, these systems allow participants to consciously recalibrate their movements and gradually shift from cognitively guided steps to more automated patterns as training advances (Pelosin et al., 2022; Pullia et al., 2023).

Immersive and semi-immersive VR systems extend this modality by incorporating obstacles, pathways, or DT elements that require visuomotor coordination and postural adjustments, engaging both cognitive and sensory systems in gait regulation (Mirelman et al., 2013). These environments are especially effective in addressing instability, reduced adaptability, and fall risk, features strongly linked to impaired visuospatial integration in PD (Keshner, 2004).

1.11.2 Auditory, Haptic, and Multimodal Feedback for Temporal and Sensorimotor Regulation

While visual cues mainly influence the spatial aspects of gait, auditory BF focuses on temporal regulation. Rhythmic cues or real-time cadence displays can support step timing and consistency, engaging rhythm-generation mechanisms that are often impaired in PD (Schlick et al., 2012). By externally reinforcing gait rhythm, auditory feedback may reduce double-support duration and enhance temporal symmetry, particularly in individuals with marked bradykinetic gait patterns (Rochester et al., 2004; Thaut et al., 1996). Haptic BF, although less frequently employed in treadmill-based PD rehabilitation, offers tactile information through vibration or pressure patterns delivered to the feet, shins, or trunk. These cues can improve proprioceptive awareness and facilitate foot-clearance adjustments or step initiation, areas often affected in PD due to impaired sensorimotor integration (Novak and Novak, 2016). Although the evidence is still preliminary, haptic feedback systems may be especially beneficial for individuals who experience visual overload or require discreet cues during DT walking. Multimodal systems combine visual, auditory, and haptic feedback to simultaneously target both the spatial and temporal components of gait. Emerging research suggests that integrating multiple sensory modalities can reinforce sensorimotor integration more effectively than single-modality approaches (Pelosin et al., 2022; Rudinskiy et al., 2014). For instance, systems that pair visual step-length cues with rhythmic auditory guidance may produce synergistic effects, encouraging both amplitude regulation and rhythmic consistency during treadmill walking (Ginis et al., 2016). Overall, the variety of BF methods reflects the multifaceted nature of gait impairment in PD. Each modality targets specific physiological processes, spatial control, rhythm generation, and proprioceptive accuracy and can be chosen strategically based on the patient’s particular deficits and training objectives.

1.12 Effects of Biofeedback on Spatiotemporal Gait Parameters

The use of BF during treadmill training has shown promising potential in modulating several gait parameters that are typically impaired in individuals with PD. These parameters, including step length, gait variability, propulsive force, cadence, and postural control, represent core components of locomotor function that often deteriorate due to impaired automaticity and altered sensorimotor integration. Evidence from feasibility trials, VR-enhanced treadmill protocols, and single-session studies suggests that real-time sensory information allows individuals with PD to correct gait deviations more effectively and engage residual mechanisms of motor learning (Rudinskiy et al., 2014; Schlick et al., 2012). As detailed below, different types of feedback selectively influence spatial, temporal, variability-related, and postural aspects of gait, providing a nuanced picture of how BF-based interventions may restore more efficient and stable locomotor patterns.

1.12.1 Spatial Parameters: Step Length, Stride Length, and Propulsive Force

Spatial gait parameters are highly responsive to visual and VR-based feedback. Studies using visual cues like step-length adjustments or projected foot trajectories have shown notable improvements in step amplitude, often within the first training session (Pullia et al., 2023; Schlick et al., 2012). These improvements are due to the heightened salience of spatial information and the ability of visual feedback to compensate for proprioceptive underestimation, a common issue in people with PD who experience gait difficulties. Treadmill-based visual cueing also enhances stride length consistency and lateral foot placement, indicating better gait symmetry and spatial stability (Rudinskiy et al., 2014). More recent research examining force-related feedback has demonstrated that visualising propulsive force during walking allows participants to increase anterior ground reaction forces, supporting more efficient forward movement (Baudendistel et al., 2024). This is particularly relevant for addressing festinating gait and reduced propulsion, which greatly impact mobility in PD.

1.12.2 Temporal Parameters: Cadence, Timing, and Rhythm Regulation

Temporal aspects of gait, including cadence and step timing, are often impaired in PD due to deficits in internal rhythm generation. BF targeting timing parameters, whether through auditory cues or real-time cadence displays, helps reinforce rhythmic consistency and reduce temporal variability (Schlick et al., 2012). Studies combining treadmill walking with rhythmic cueing have shown improvements in temporal symmetry and decreases in the duration of the double-support phase, indicating a more stable gait pattern (Rochester et al., 2004; Thaut et al., 1996). In VR-based protocols, temporal improvements often occur alongside better DT performance, suggesting that rhythmic stability may also benefit from cognitive engagement during feedback training (Mirelman et al., 2013; Pelosin et al., 2022). These findings are consistent with the view that externally supported rhythm generation can partially compensate for basal ganglia dysfunction and facilitate the transition from cognitively guided timing to more automated rhythmic control.

1.12.3 Variability, Asymmetry, and Dynamic Stability

Gait variability and asymmetry are considered sensitive indicators of fall risk in PD. BF interventions have shown reductions in step-to-step variability and improvements in mediolateral stability (Pelosin et al., 2022; Pullia et al., 2023). These results indicate that feedback not only affects specific gait parameters but may also enhance dynamic balance control through better motor-cognitive interaction. Evidence from motion-capture and IMU-assisted treadmill protocols suggests that improvements in variability may stem from improved integration of sensory cues with motor planning, a process that tends to deteriorate as PD progresses (Hausdorff, 2009). By fostering consistency and accuracy in foot placement, BF helps stabilise gait patterns that might otherwise be prone to disruption, especially in multitasking or visually demanding settings. Such changes align with the notion that sensorimotor recalibration and enhanced attentional engagement can reduce the “noisy” expression of gait control typically seen in PD.

1.12.4 Postural Orientation and Trunk Control

Postural alignment and trunk movement are vital for maintaining balance and efficient walking. Feedback that offers real-time information on trunk angle or centre-of-mass displacement has demonstrated positive effects on upright posture, trunk rotation, and weight-shifting control (Rudinskiy et al., 2014). These adjustments are likely driven by increased awareness of postural deviations and the activation of compensatory pathways that support axial stability. VR-enhanced treadmill systems seem particularly effective in challenging and improving postural control through tasks that require head-trunk coordination and anticipatory balance adjustments (Mirelman et al., 2016b). Improvements in these areas may indirectly contribute to the reduction in fall frequency seen in feedback-based treadmill training programmes. By reinforcing axial mobility and postural responsiveness, domains often resistant to pharmacological treatment, BF-based interventions target a crucial component of safe mobility in PD.

Taken together, the observed changes in spatial, temporal, variability-related, and postural parameters suggest more than isolated, task-specific adaptations. They are consistent with enhanced gait automaticity, improved sensorimotor integration, and more efficient motor-cognitive resource allocation, in line with the theoretical framework outlined in Sections 1.4 and 1.9 and the neurophysiological mechanisms discussed in Section 1.14. Rather than merely “normalising” individual gait metrics, BF-based treadmill training appears to support a more robust and flexible control of locomotion under changing task demands, providing a mechanistic bridge between the kinematic improvements described here and the clinical benefits in falls, balance, and functional mobility reported in Section 1.12.

Overall, BF has demonstrated the potential to enhance a wide range of gait parameters in PD. Spatial features, such as step length and propulsion, respond consistently and quickly. Temporal parameters benefit from rhythmic support. In more complex areas, variability, asymmetry, and postural stability improve through increased sensory and cognitive engagement. Although the magnitude of these effects varies across studies, the accumulated evidence highlights BF as a promising adjunct for restoring gait function through targeted, parameter-specific motor learning.

1.13 Clinical Effects of Treadmill Training with Biofeedback: Falls, Balance, and Functional Mobility

The clinical importance of treadmill training enhanced with BF in PD is increasingly recognised, especially for outcomes related to fall reduction, balance control, and functional mobility. These areas are vital in PD rehabilitation, as motor automaticity declines and individuals become more dependent on cognitive and sensory compensatory mechanisms to walk safely. Adding BF to treadmill exercise seems to boost its clinical benefit by improving patients' ability to manage their gait, respond to postural challenges, and maintain mobility in daily life (Schlick et al., 2012; Studer et al., 2017). Evidence indicates that clinical improvements come from the combination of repetitive motor practice, focused attention, and the corrective feedback that supports motor learning, which remains partly functional despite basal ganglia issues (Jellish et al., 2015; Pullia et al., 2023).

1.13.1 Fall Reduction and Improvement of Dynamic Stability

Falling is one of the most functionally disabling symptoms in PD and is closely linked to gait variability, difficulties in overstepping obstacles, and deficits in DT performance. BF-enhanced treadmill training has been shown to significantly reduce the frequency of falls. Multicentre trials using obstacle-rich VR environments showed a marked decrease in fall rates compared to treadmill training alone (Mirelman et al., 2016b). These results suggest that immersive feedback helps promote safer motor behaviour by enhancing visuomotor responses, anticipatory adjustments, and sensorimotor integration in complex walking conditions.

The clinical importance of fall prevention extends beyond immediate safety: fewer falls lead to increased confidence in mobility, less fear of falling, and greater participation in daily activities (Venhovens et al., 2020). The clinical improvements observed highlight the potential of BF to oppose the progressive destabilisation seen in PD, thereby supporting long-term independence.

1.13.2 Effects on Balance Control and Postural Regulation

Balance impairments in PD originate from deficits in axial control, impaired sensory integration, and diminished automatic postural responses. BF systems providing real-time information on trunk alignment, foot placement, or centre-of-mass position assist participants in modulating postural behaviour during treadmill walking (Rudinskiy et al., 2014). These sensory cues improve awareness of postural deviations and encourage corrective strategies that enhance both static and dynamic balance.

Studies using VR-enhanced treadmill paradigms show that repeated exposure to visuospatial and motor-cognitive challenges supports postural adaptability and improves clinical measures such as the Berg Balance Scale and Mini-BESTest (Pelosin et al., 2022; Cornelia Schlick et al., 2012). Such balance improvements are likely due to the combined effects of increased attentional focus, heightened sensory salience, and repeated practice of postural adjustments under structured conditions.

Additionally, enhancements in trunk rotation and weight-shifting control have been observed following feedback-guided training, supporting the idea that axial mobility may benefit from external sensory guidance (King and Horak, 2011). These effects are especially relevant, given that axial rigidity often resists pharmacological treatment.

1.13.3 Functional Mobility: Walking Speed, Endurance, and Dual-Task Capacity

Functional mobility, including walking speed, gait endurance, and the ability to maintain locomotor function during cognitively demanding tasks, is a vital outcome in PD rehabilitation. BF used during treadmill walking has been linked to improvements in walking speed, increased stride amplitude, and better rhythmicity (Pullia et al., 2023; Cornelia Schlick et al., 2012). These gains demonstrate improved gait efficiency and greater confidence in stepping, both of which contribute to more effective mobility in daily life. Several studies also report improvements in endurance measures, such as the 6-Minute Walking Test, following treadmill-based interventions that incorporate visual feedback (Pelosin et al., 2022). The combined aerobic and motor-learning approach seems to support cardiovascular conditioning and locomotor capacity concurrently. Importantly, motor-cognitive treadmill training, including visual feedback and VR-based obstacle overcoming, has shown beneficial effects on DT walking (Mirelman et al., 2013). This is clinically relevant because DT ability is strongly connected to real-world mobility and fall risk. By challenging both motor and executive functions, programs that use BF may strengthen cognitive control of gait and enhance adaptability in dynamic environments (Yogev-Seligmann et al., 2008).

BF-based treadmill training exerts measurable clinical benefits across several domains central to PD management. Improvements in fall reduction, balance control, and functional mobility appear to arise from enhanced motor learning, improved sensorimotor integration, and strengthened cognitive regulation of gait. While the magnitude of these effects varies with the type of intervention and patient characteristics, the accumulated evidence supports the integration of BF as an effective clinical tool for targeting complex gait and balance deficits in PD.

1.14 Intervention Parameters: Dose, Intensity, and Duration

The definition of suitable training parameters, including dose, intensity, and duration, is crucial to maximise the therapeutic benefits of treadmill training combined with BF in PD. As with other types of exercise therapy, these factors influence the extent and durability of motor learning, the use of cognitive resources, and the overall physiological response to training (Pelosin et al., 2022; Pullia et al., 2023). Although the literature on BF-specific prescriptions remains limited, several principles can be drawn from studies using sensorized treadmills, VR-enhanced environments, and closed-loop feedback systems (Schlick et al., 2012; Studer et al., 2017) . Subsequently, the fundamental elements of dosage, intensity, and timing are summarised.

1.14.1 Training Frequency and Total Dose

Evidence from treadmill-based rehabilitation indicates that a minimum frequency of two to three sessions per week is necessary to produce meaningful improvements in gait performance, especially when the intervention targets gait amplitude, propulsion, or dynamic stability (Pelosin et al., 2022). Interventions using VR-enhanced treadmill training or intensive visual feedback often incorporate higher training frequencies and longer programmes, ranging from 4 to 12 weeks, to ensure sufficient repetition and reinforcement of improved gait patterns (Mirelman et al., 2016a). The training dose closely aligns with the principles of motor learning, as repeated exposure to corrective feedback facilitates the consolidation of more effective motor strategies. High-volume gait practice, facilitated by treadmill walking, supports this consolidation, while adding feedback stimulates the cognitive phase of learning and improves retention over time (Abbruzzese et al., 2016). Although the exact dose-response relationships remain to be clarified, longer and more frequent programmes tend to generate more stable improvements, particularly in complex outcomes such as balance, variability, and DT gait.

1.14.2 Training Intensity: Aerobic Load and Cognitive Engagement

Intensity reflects a multidimensional aspect of the intervention, influenced by factors such as treadmill speed, incline, cognitive load, and the complexity of feedback. In many studies, treadmill speed is adjusted to the participant’s comfort level to ensure safety and promote adherence; however, gradually increasing speed may stimulate greater step amplitude, higher propulsive forces, and improved rhythmicity (Pullia et al., 2023; Schlick et al., 2012). From a metabolic standpoint, treadmill training can support moderate-intensity aerobic exercise, which is associated with improved endurance and gait efficiency in PD (Johansson and Jarnlo, 2020). When BF is incorporated, intensity also involves cognitive demands. VR-based treadmill paradigms, for example, introduce visuospatial and executive challenges in successfully crossing obstacles or incorporating DT elements (Mirelman et al., 2016a). These components increase attentional engagement, encourage adaptive motor control, and may contribute to broader functional improvements. Crucially, the combination of physical and cognitive load must be carefully calibrated to prevent excessive fatigue or deterioration in movement quality, especially in individuals with advanced PD or coexisting cognitive impairment.

1.14.3 Duration of Sessions and Structure of Training Progression

Most treadmill and BF protocols use session durations of 20 to 45 minutes, divided into continuous or interval walking bouts, depending on fatigue levels and the complexity of the feedback tasks (Studer et al., 2017). Longer sessions may offer more opportunities to practice corrected gait patterns; however, they must be balanced against the rapid fatigability typically associated with PD. Progressive adjustment of training difficulty is another key aspect of duration. In VR-enhanced protocols, this progression may involve increasing obstacle density, narrowing walking paths, or introducing DT challenges. In visual BF systems, progression can involve gradually decreasing the feedback frequency to promote internalisation and support the transition toward more automatic gait control (Sigrist et al., 2013). Sustained improvements observed in follow-up assessments of VR-based treadmill training highlight the importance of repeated exposure to challenging yet achievable training stimuli (Pelosin et al., 2022). Programmes lasting 8-12 weeks generally produce larger and more durable effects compared with shorter interventions.

The clinical and motor learning benefits of treadmill training with BF rely on the careful adjustment of training parameters. Adequate frequency promotes repetition-based learning; intensity influences both physical and cognitive involvement; and appropriately structured duration with progressive challenges supports consolidation. Although specific guidelines tailored to BF interventions are still being developed, current evidence supports moderate to high repetition, increasing difficulty, and sustained programmes to achieve meaningful improvements in gait and mobility in PD.

1.15 Neurophysiological Mechanisms Underpinning Biofeedback-Enhanced Treadmill Training

The beneficial effects of treadmill training with BF in PD are supported by several neurophysiological mechanisms that compensate for impaired automaticity, enhance sensory integration, and promote motor learning. In PD, gait and balance deficits mainly originate from basal ganglia dysfunction, which disrupts the internal generation and scaling of movement, forcing individuals to rely on attentional and cortical compensatory strategies. By providing continuous external information during gait, BF appears to facilitate these compensatory mechanisms by increasing the salience of sensory cues and activating alternative neural pathways that support locomotor control (Jellish et al., 2015; Schlick et al., 2012). One of the key mechanisms proposed involves the recruitment of cortico-cerebellar networks to compensate for impaired basal ganglia-thalamocortical loops. Visual and auditory feedback delivered during treadmill walking may encourage the shift from dysfunctional internal cues towards externally driven motor control, enabling patients to regulate step length, rhythmicity, and posture more effectively (Morris, 2010; Nieuwboer et al., 2007). This compensatory engagement supports the cognitive phase of motor learning, allowing individuals to consciously correct gait deviations before progressive repetition fosters partial re-automatization. Another mechanism pertains to the enhancement of sensorimotor integration. PD is linked with impaired proprioceptive processing and inaccurate perception of limb position and movement amplitude. Real-time visual feedback, whether through step targets, kinematic overlays, or VR-based obstacle cues, provides highly salient sensory input that recalibrates distorted internal models of gait (Pullia et al., 2023; Rudinskiy et al., 2014). By aligning visual information with proprioceptive feedback during repeated gait cycles, BF may strengthen the integration of multimodal sensory signals and reduce dependence on compromised basal ganglia circuits (Konczak et al., 2009). BF also engages attentional and executive resources, which are essential for modulating gait in PD. VR-based treadmill protocols require participants to perform visuospatial planning, avoid obstacles, and task switching (Mirelman et al., 2016a). This cognitive engagement has been associated with improvements in DT walking and may reflect enhanced recruitment of frontoparietal networks responsible for top-down control of movement (Yogev-Seligmann et al., 2008). The interaction between cognitive load and motor performance supports the idea that BF-enhanced training strengthens the motor-cognitive link, which is crucial for approaching real-world. A further mechanism involves implicit and explicit motor learning processes. Continuous feedback supports explicit learning by providing immediate information about performance errors, while the rhythmic and repetitive nature of treadmill walking promotes the implicit consolidation of corrected movement strategies (Pelosin et al., 2022; Studer et al., 2017). Over time, reduced reliance on feedback may indicate a gradual internalisation of improved gait patterns. Finally, visual feedback may boost motivational drive and task engagement, factors known to influence dopamine release and plasticity within surviving neural circuits (de la Fuente-Fernández and Stoessl, 2001). Increased motivation and reward expectation can enhance motor output and facilitate learning, making BF not merely a corrective tool but also a modulator of neuroplastic potential.

1.16 Research Gaps and Future Directions for Training with Biofeedback

The literature on treadmill training combined with BF in PD has expanded considerably, yet several limitations continue to impact the interpretation and applicability of existing findings. A primary concern involves the variability of intervention protocols, especially regarding the type of feedback provided (visual, auditory, VR-based), the targeted parameters, and the structure of training sessions. This inconsistency complicates the identification of which components are most effective and hinders comparisons across studies (Pullia et al., 2023; Schlick et al., 2012). Furthermore, inconsistent reporting of technical details, such as feedback latency, display modality, or sensor setup, diminishes reproducibility (Sigrist et al., 2013). Another limitation pertains to the small sample sizes common in many feasibility and pilot studies. These often lack randomisation or suitable control conditions, reducing statistical power and increasing the risk of exaggerated effect estimates (Rudinskiy et al., 2014; Studer et al., 2017). As noted in broader methodological reviews, small samples can compromise reliability and hinder the identification of significant clinical subgroups (Button et al., 2013). A further recurring issue is the brief duration of follow-up. Although improvements in gait and balance are often observed post-intervention immediately, few studies examine long-term retention beyond 3-6 months (Mirelman et al., 2016b). Without extended data, it remains uncertain whether these effects represent enduring motor learning or short-term compensatory changes. This concern is especially relevant for outcomes related to fall risk and DT capacity, which depend on sustained gains over time (Keus et al., 2014). Lastly, research is limited by variability in outcome measures and the prevalent use of controlled laboratory environments. Few studies explore BF-based treadmill training in home or community settings, where ecological validity and adherence are crucial (Pelosin et al., 2022; Pullia et al., 2023). This gap restricts understanding of how improvements translate into real-world mobility and daily functioning. Overall, while current evidence appears promising, future research must adopt more standardised protocols, involve larger and adequately powered samples, and include longer follow-up periods to clarify the genuine clinical benefits of feedback-enhanced treadmill training in PD.

1.17 Overall Conclusions and Integration of Assessment and Training Perspectives

Although current evidence highlights the potential of treadmill training with BF to enhance gait performance in PD, several research avenues remain essential to consolidate findings and align rehabilitation strategies with the need for sensitive assessment tools and targeted interventions. Future research should focus on refining methods, developing advanced DT paradigms, and integrating aerobic and cognitive demands within personalised training protocols.

A key research priority involves early detection of motor-cognitive impairments, especially in individuals with mild or newly diagnosed PD. Comparative studies between early-stage patients and healthy controls are needed to clarify how cognitive load affects gait at different disease stages and whether DT performance can serve as a biomarker for functional decline (Pelosin et al., 2022; Pullia et al., 2023). This research is particularly pertinent because subtle deficits in automaticity may precede the onset of obvious motor symptoms, providing a window of opportunity for early intervention (Lord et al., 2010).

Another important direction is to optimise aerobic treadmill training protocols enriched with BF. Future studies should systematically assess how training intensity, measured through metabolic indicators such as oxygen consumption ($\dot{V}O_2$), interacts with improvements in gait speed, step length, and locomotor stability. Understanding dose-response relationships will help define the metabolic thresholds necessary to induce neuroplastic changes, thereby improving both gait and cardiorespiratory fitness (Mirelman et al., 2016b; Pelosin et al., 2022). The incorporation of real-time physiological monitoring (e.g., heart rate, $\dot{V}O_2$ estimates) into treadmill feedback systems offers an innovative opportunity to tailor training intensity to individual capacities (Johansson and Jarnlo, 2020).

Ultimately, developing adaptive BF systems capable of modulating task difficulty in real-time based on performance presents a promising technological advance. Such systems could dynamically adjust step targets and cognitive demands to ensure each participant trains at an optimal challenge point, maximising motor learning and retention (Sigrist et al., 2013). This approach aligns with principles of precision rehabilitation and may facilitate the translation of laboratory protocols into scalable clinical tools.

In conclusion, future research should aim at integrating standardised DT assessments, metabolic monitoring, and adaptive feedback technologies to support evidence-based, personalised rehabilitation strategies that address both motor and cognitive contributors to gait dysfunction in PD.

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CHAPTER 2: Experimental Aims

2.1 Purposes and Research Questions and General Framework

Gait in PD is characterised by reduced automaticity, increased reliance on attentional resources, and significant vulnerability to DT conditions. Every day, walking rarely occurs in isolation: individuals must continuously integrate cognitive demands, such as planning, environmental monitoring, or performing concurrent tasks, while regulating locomotor output. Impairments in this motor-cognitive integration are among the earliest and most disabling features of PD.

Despite growing clinical interest, research on DT gait still faces variability in cognitive tasks, inconsistent methodological approaches, and limited understanding of how cognitive load interacts with gait control in early PD. At the same time, although treadmill-based interventions with BF have shown promise for improving gait performance, the specific role of attentional demands, contextual interference, and exercise intensity in driving motor improvements remains insufficiently understood.

This thesis addresses these gaps by integrating methodological development, mechanistic investigation, and interventional research into a single programme, structured across three complementary studies.

2.2 Study 1 - Development and Validation of a Graded Dual-Task Gait Test in Healthy Older Adults

Walking requires the ongoing coordination of cognitive and motor resources. When attentional or executive demands increase, gait performance in older adults usually shows slower speed, shorter steps, and longer double-support time. However, current DT paradigms are methodologically inconsistent, as tasks vary in difficulty, domain, and pacing, making it difficult to compare findings across studies or to determine dose-response relationships between cognitive load and gait decline.

The purpose of Study 1 was to develop and validate a standardised DT gait test featuring three controlled levels of cognitive difficulty (serial subtraction by -1, -3, -7).

This graded protocol was designed to:

- Systematically manipulate cognitive load.
- Quantify its effects on spatiotemporal gait parameters in a reproducible manner.
- Measure DT costs across different conditions.
- Investigate whether a difficulty-dependent, linear decline in gait occurs with increasing cognitive load.
- Establish normative data and validate a continuous measure of cognitive load (Difficulty Index).

Study 1 hypothesised that:

- Increasing cognitive difficulty would lead to progressive deterioration in gait, such as slower speed, shorter steps, and increased double support.
- DTC would vary across DT-1, DT-3, and DT-7, reflecting greater interference.
- The Difficulty Index would demonstrate a strong load-response relationship with gait parameters.
- Older adults would display consistent within-individual load-response patterns, indicating stable motor-cognitive coupling.

2.3 Study 2 - Comparing Dual-Task Gait Responses Between Healthy Older Adults and Individuals with Early-Stage Parkinson's Disease

In early PD, gait deficits often appear before obvious disability and may worsen under DT conditions. Understanding how cognitive load influences gait in early stages is vital for detecting subtle functional vulnerabilities, enhancing diagnostic tools, and guiding personalised rehabilitation strategies.

Study 2 expanded the graded DT paradigm from Study 1 to compare individuals with early-stage PD and age-matched healthy controls. The aim was to examine:

- Whether PD already exhibits reduced gait automaticity during single-task walking.
- Whether increasing cognitive load causes a more pronounced deterioration in PD compared to controls.
- Whether DTC differ between groups and load levels.
- Whether PD and controls differ in the slopes of the load-response relationships captured by the Difficulty Index.
- Whether clinically meaningful functional thresholds (e.g., walking $< 1.0\ m \cdot s^{-1}$) are easier for individuals with PD to cross.

Study 2 hypothesised that:

- Participants with PD walk more slowly, with shorter steps and longer double support, compared to controls, even during single-task conditions.
- Both groups show load-dependent gait deterioration, but PD exhibits higher DTC across all levels.

- The Difficulty Index captures group differences in baseline performance while indicating similar load sensitivity within individuals.
- DT gait serves as a sensitive marker of early functional impairment in PD.

Study 2, therefore, aimed to characterise how early PD alters motor-cognitive integration and to evaluate whether the graded DT test can serve as a robust clinical tool for early detection and monitoring.

2.4 Study 3 - Effects of Treadmill Biofeedback Training and Exercise Intensity on Gait and Aerobic Adaptation in Parkinson's Disease

Real-time BF during treadmill walking has emerged as a promising strategy to enhance step length, cadence, and locomotor stability in PD by increasing attentional engagement and promoting error-based motor learning. Yet, several critical questions remain unanswered:

- Does adding cognitive feedback (such as step-length and cadence targets) improve gait more than traditional treadmill training?
- Does high contextual interference, involving random alternation between feedback cues, enhance learning by increasing attentional load?
- Does higher metabolic intensity amplify training effects?
- How do training-related improvements manifest under DT conditions, where gait is usually most vulnerable in PD?

Study 3 investigated the effects of a 4-week treadmill-based intervention, comparing:

- High-interference biofeedback training (BF_HI_T)
- Low-interference biofeedback training (BF_T)
- Traditional treadmill training (T_T)
- High-interference biofeedback combined with higher metabolic intensity (BF_HI_HM_T)
- Control group

The study assessed outcomes under both single- and DT gait conditions, combining spatiotemporal metrics and aerobic parameters (including $\dot{V}O_2$, ventilatory thresholds, and exercise intensity).
Study 3 hypothesised that:

- BF training, particularly under high interference, yields greater improvements in step length and double support compared to traditional treadmill training.

- The effects would be most noticeable in conditions with high cognitive load, where gait faces the greatest challenge.
- Increased aerobic intensity would further enhance gait outcomes and may also improve aerobic fitness.
- Gains in gait parameters would be linked to gradual decreases in the Difficulty Index.

The purpose of Study 3 was therefore to examine whether attentional engagement, contextual interference, and metabolic intensity are crucial factors influencing gait improvement in PD, thereby identifying mechanisms that could inform more effective, personalised rehabilitation strategies.

2.5 Overall Research Questions

Across the three studies, this thesis aimed to answer the following overarching research questions:

- Can DT gait assessment be standardised through a graded difficulty protocol that can produce reproducible, load-dependent deterioration in gait performance?
- Does early-stage PD alter the relationship between cognitive load and gait control, and can a graded DT paradigm sensitively detect early deficits in gait automaticity and motor-cognitive integration?
- Can treadmill-based BF training, especially when delivered with high contextual interference and increased metabolic intensity, improve spatiotemporal gait parameters and resilience to cognitive load in individuals with PD?
- How are training-related changes reflected in single-task and DT conditions, and what is the role of exercise intensity and attentional load in promoting gait improvements?

Together, these questions underpin the scientific rationale for the thesis, which aims to develop reliable assessment tools and identify effective, mechanism-informed rehabilitation strategies that can enhance gait automaticity, stability, and DT performance in individuals with PD.

CHAPTER 3: Graded Dual-Task Walking in Healthy Older Adults: A Dose-Response Relationship between Cognitive Load and Gait Deterioration

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Abstract

Graded dual-task (DT) walking assessments may reveal how increasing cognitive load interferes with gait in older adults. We examined dose-response relationships between cognitive difficulty and gait in 30 healthy older adults (mean age, 69.6 years). Participants walked overground under single-task (ST) and three serial-subtraction DT conditions (DT-1, DT-3, DT-7). Spatiotemporal parameters were recorded with an optical system, and dual-task cost (DTC) and a Difficulty Index (DI) were calculated. Repeated-measures ANOVA was used to compare the conditions, while linear regression was used to quantify the relationships between DI, gait, and fatigue ratings. Gait speed, step length, double support, and cadence differed across conditions (all $p < 0.001$), with speed DTC increasing from approximately 6% in DT-1 to over 20% in DT-7. DI increased from 0.66 ± 0.20 (DT-1) to 3.45 ± 2.85 (DT-7; $p < 0.001$), while correct answers decreased from 12.0 ± 2.97 to 3.97 ± 1.65 ($p < 0.001$). In DT-3 and DT-7, 7 and 8 participants respectively walked below 1.0 $m \cdot s^{-1}$. DI correlated strongly with speed and step length and positively with double support. Mental, but not physical, fatigue increased with task difficulty. These findings support a graded, approximately linear dose-response relationship between cognitive load and gait deterioration, highlighting the clinical importance of standardised multi-level DT protocols.

3.1 Background on Dual-Task Walking Tests

Dual-task (DT) walking tests are commonly used to assess how well individuals sustain gait performance while engaged in an additional cognitive or motor task. These evaluations are vital because walking in everyday life is seldom performed alone; it is often accompanied by other activities, such as conversing, avoiding obstacles, or planning future actions (Al-Yahya et al., 2011; Woollacott and Shumway-Cook, 2002). From a neurophysiological perspective, gait control relies on the interaction between motor and cognitive processes, especially executive function and attention. When attentional demands surpass available resources, performance declines known as dual-task costs (DTC), affecting gait, the cognitive task, or both (Li and Lindenberger, 2002; Yogev-Seligmann et al., 2008a). These changes are more evident in older adults and populations with neurological impairments, and DT performance has been suggested as a sensitive marker of early cognitive decline and impaired motor-cognitive integration (Al-Yahya et al., 2011; Verghese et al., 2007).

The value of this type of test is further supported by a recent review within the preclinical spectrum of cognitive ageing, in which subjective cognitive decline (SCD) describes self-reported worsening of cognitive function without objective deficits on standard tests. At this at-risk stage, DT walking functions as an ecological stress test of motor-cognitive integration: by engaging executive attentional resources during walking, it can detect subtle gait signatures, such as slower speed, shorter step length, and increased double support/variability compared to single-task conditions, which may not be apparent under simpler circumstances. Converging evidence suggests that these changes tend to increase with the level of cognitive load, endorsing the use of standardised DT protocols with progressively higher difficulty for early vulnerability detection, risk assessment, and long-term monitoring (Salzman et al., 2025).

3.1.1 Current Research Gaps

Despite the increasing use of DT walking tests in clinical and research settings, standardised protocols are still lacking. The significant variation in secondary tasks, including differences in domain (e.g., mental arithmetic, verbal fluency, reaction time), stimulus pacing, step sizes in subtraction, and response modality, leads to non-comparable attentional demands and task-specific interference patterns, making direct comparisons within and between individuals difficult (Goh et al., 2021). Meanwhile, procedural inconsistencies, such as varying instructions regarding task prioritisation and changes in procedures between pre- and post-assessments, further complicate the interpretation of DT effects. Outcome reporting is also inconsistent: gait speed is often reported, whereas

spatiotemporal and variability measures are included less frequently; even the formula used to calculate DTC varies across studies (Goh et al., 2021).
Furthermore, performance on the non-gait task is often not reported in either single or DT conditions, which prevents the calculation of cognitive-task DTC and restricts conclusions about attentional allocation. Lastly, small sample sizes and limited longitudinal follow-up slow down the development of normative references and the assessment of changes over time (Plummer et al., 2015).
Taken together, these gaps highlight the need for protocols that clearly define difficulty levels and make them comparable within the same set of tasks, apply consistent instructions, report outcomes for both gait and cognitive tasks, and adopt standardised DTC definitions to enhance comparability and clinical usefulness (Goh et al., 2021).

3.1.2 Development of a DT Walking Test with Three Levels of Difficulty

The present study introduces a standardised DT walking paradigm to address these limitations. The test includes three levels of increasing difficulty in a serial subtraction task: counting backwards by ones (DT-1), by threes (DT-3), and by sevens (DT-7). Serial subtraction is especially relevant, as it involves working memory, attention, and executive function control (Dubost et al., 2006). By systematically adjusting task difficulty, this protocol aims to simulate real-life multitasking demands and to determine whether threshold points can be identified where increasing cognitive load leads to deterioration in gait parameters, either as higher DT costs or as absolute declines.

3.1.3 Study Rationale and Utility

Everyday walking constantly competes with cognitive demands, making DT walking a practical measure of cognitive-motor integration. Establishing baseline performance in healthy older adults offers normative references to identify deviations caused by ageing, fatigue, or neurodegenerative processes. As cognitive load increases, individuals often adopt compensatory strategies, such as reallocating attention to the secondary task and maintaining gait stability by slowing down, shortening steps, or increasing double support, though the extent and pattern of these adjustments vary depending on the nature and difficulty of the secondary task (Goh et al., 2021). Evidence from preclinical populations further indicates that DT testing can detect subtle declines in spatiotemporal parameters even when standard neuropsychological performance stays within normal limits, supporting its role in early detection of cognitive-motor vulnerability (Salzman et al., 2025).
In this context, our study uses a single-task condition along with three subtraction tasks of increasing difficulty (DT-1, DT-3, DT-7) to measure absolute spatiotemporal parameters and their DTC

compared to single-task walking. By linking changes in gait to increasing cognitive demands within a standardised framework, the protocol aims to produce normative references, improve interpretability within and across studies, and deliver clinically relevant information, such as whether walking speed approaches commonly cited safety thresholds under cognitive load, which may aid in risk stratification, long-term monitoring, and assessing rehabilitation outcomes (Goh et al., 2021; Salzman et al., 2025).

3.1.4 Broader Applications

A multi-intensity DT walking assessment can be used across clinical populations to track disease progression, measure cognitive-motor interference, and assist in falling risk stratification. Linking changes in spatiotemporal parameters and DTC to increasing cognitive demands uncovers load-response patterns that are not visible during single-task conditions and that depend on the type and severity of the secondary task (Almutairi et al., 2025; Al-Yahya et al., 2011; Goh et al., 2021).

From a clinical perspective, monitoring whether walking speed approaches or exceeds safety-relevant thresholds under higher loads, along with increases in double support, provides actionable signals associated with reduced stability and potential falls (Adam et al., 2023; Montero-Odasso et al., 2012). Standardised multi-intensity protocols also enable comparison across populations and settings and support the derivation of reference curves and cut-points, beyond which increasing cognitive load consistently leads to deterioration in gait, either as high DTC or as absolute decrements, thus improving comparability and clinical utility (Goh et al., 2021; Plummer et al., 2015). Moreover, emerging index-based approaches (e.g., Performance Index; automaticity/normalcy index) demonstrate how composite metrics can summarise motor-cognitive coupling and may support anchoring such cut-points in populations at risk or within specific disease contexts (Brauner et al., 2021; Liu et al., 2023; Meng et al., 2025). Finally, evidence that alterations are detectable even at preclinical stages reinforces the use of DT testing for early detection and ongoing monitoring of cognitive-motor vulnerability (Salzman et al., 2025).

3.1.5 Study Aim and Experimental Hypotheses

The initial premise was that as the concurrent cognitive task becomes more challenging, more attention is diverted from walking. Therefore, interference with gait should increase rather than stay uniform across tasks. Accordingly, we anticipated that the DTC would vary across difficulty levels, with greater costs at higher subtraction loads (DT-1 < DT-3 < DT-7). This expectation stems from the idea that as executive-attentional resources are allocated to the cognitive task, fewer remain for gait control, leading to measurable declines in key spatiotemporal parameters (e.g., slower speed and shorter steps, with compensatory increases in double support). We also anticipated this pattern to be consistent within individuals, not just in group averages. For each participant, the DTC should increase in an approximately linear manner as task difficulty rises, producing positive individual slopes when cost is regressed against difficulty level. Practically, the same person tested under increasingly difficult subtraction demands should show a graded deterioration relative to single-task baselines, characterised by rising costs for speed, step length, and cadence, as well as an increased reliance on stability (double support). This individual-level linearity provides a rigorous test of the hypothesis that cognitive load and gait cost are directly linked, rather than merely differing between conditions at the group level.

3.2 Materials and Methods

3.2.1 Participants

Thirty healthy older adults (15 males, 15 females; mean age ± SD, 69.6 ± 5.2 years) participated voluntarily in the study. Participants were recruited from the local community through flyers, advertisements, and word of mouth. Inclusion criteria included: no orthopaedic disorders, no current or recent neurological symptoms (within the past six months), no pain in either lower extremity, and no evidence of cognitive impairment, as indicated by a Mini-Mental State Examination (MMSE) score > 24 (Folstein et al., 1975). Short Physical Performance Battery (SPPB); scores below 10/12 were considered indicative of mobility limitation, and such participants were excluded (Guralnik et al., 1994). Perceived health status was assessed using the Short Form Health Survey (SF-36), from which norm-based Physical Component Summary (PCS) and Mental Component Summary (MCS) scores were derived (mean = 50, SD = 10). For screening purposes, a PCS or MCS score below 40 (more than 1 SD below the population mean) was considered below normal limits and resulted in exclusion (Apolone and Mosconi, 1998; Garratt and Stavem, 2017). Psychological variables were assessed using the State-Trait Anxiety Inventory Trait form (STAI-Y II) and the Beck Depression Inventory-II (BDI-II). Participants with STAI-Y II scores of 44 or higher (clinically relevant trait anxiety) (Bieling et al., 1998; Linde et al., 2022) or BDI-II ≥ 14 (≥ mild depression) were excluded (Wang and Gorenstein, 2013). Sleep quality was assessed using the Pittsburgh Sleep Quality Index (PSQI); a global score greater than 5 indicated poor sleep quality and served as an exclusion criterion (Curcio et al., 2013).

Out of 31 individuals who volunteered to participate in the study, 30 were eligible and were included in the study. A summary of demographic and clinical variables is reported in *Table 1*.

Mean ± SD	Women	Men	Tot	*p*
N°	15	15	30	-
Age [yr]	69.3 ± 35.4	70.0 ± 6.0	69.6 ± 5.2	0.71
Weight [kg]	63.5 ± 33.2	80.6 ± 9.0	72.1 ± 13.2	<0.001
Height [cm]	160.1 ± 81.5	176.0 ± 6.2	168.0 ± 9.9	<0.001
BMI [kg.m-2]	24.8 ± 13.0	26.2 ± 3.8	25.5 ± 4.0	0.37
Years of Education [yr]	13.1 ± 6.9	13.7 ± 2.7	13.4 ± 2.7	0.55
BDI - Y II	3.3 ± 2.8	4.0 ± 5.7	3.6 ± 4.5	0.67
MMSE	27.3 ± 13.9	26.9 ± 1.2	27.1 ± 1.3	0.34
PSQI	5.6 ± 3.5	4.3 ± 2.1	5.0 ± 2.5	0.18
SPPB	12.0 ± 6.1	12.0 ± 0.0	12.0 ± 0.0	-
SF-36 PCS	51.9 ± 26.8	53.2 ± 5.8	52.5 ± 6.2	0.58
SF-36 MCS	52.9 ± 27.1	52.1 ± 6.3	52.5 ± 5.5	0.71

Table 1. Overview of demographic and clinical variables between men and women in the sample included in the study.

All participants provided written informed consent prior to enrolment. The study was conducted in accordance with the Declaration of Helsinki and approved by the University of Verona Committee for the Approval of Research on Humans (CARU; protocol no. 08/2022).

3.2.2 Study Design and Procedure

A preliminary medical evaluation, including the MMSE test, SPPB, SF-36, STAI-Y II, BDI-II, and PSQI, was performed. The study comprised two sessions. In the first session, demographic and clinical data were collected, and in the second, the experimental DT walking protocol was carried out. All sessions occurred in a temperature- and humidity-controlled exercise physiology laboratory at the University of Verona (22-25 °C, relative humidity 55-65%). Trained investigators conducted all assessments.

3.2.3 Anamnestic Session: Demographic and Clinical Data

A structured interview and validated questionnaires were used to characterise the sample. Anthropometric measures included height (168.0 ± 9.9 cm; vertical stadiometer, Seca, Leicester, UK) and weight (72.1 ± 13.2 kg; digital scale, Seca, Leicester, UK). Data on handedness and years of education were collected.

3.2.4 Experimental Session: Dual-Task Walking Protocol

Participants performed four different walking conditions in randomised order:

- Single-task (ST): walking without a concurrent task.
- Dual-task 1 (DT-1): walking while counting backwards by ones.
- Dual-task 3 (DT-3): walking while counting backwards by threes.
- Dual-task 7 (DT-7): walking while counting backwards by sevens.

Each condition was performed twice, resulting in eight trials per participant, with one-minute seated rests between trials. Walking trials took place on a 10-metre walkway. To ensure measurement of steady-state gait, participants started walking 1.5 metres before entering the sensor area and continued for 1.5 metres beyond its end. Participants were instructed to walk at a self-selected pace along a straight path.

Spatiotemporal gait parameters were divided into primary and secondary outcomes based on their clinical relevance and established sensitivity to cognitive-motor interference. Primary gait outcomes included walking speed ($m \cdot s^{-1}$), step length (cm), and double support time (% gait cycle), as they directly reflect gait efficiency and compensatory stability strategies. Secondary gait outcomes included cadence (steps/min), step length coefficient of variation (step-length CV, %), and step length asymmetry (calculated as left minus right step length, normalised to the left step length x 100), were measured using a validated photoelectric system (Witty Gate, Microgate, Bolzano, Italy) integrated with 10 one-meter photoelectric bars and dedicated software (OptoGait, Microgate, Bolzano, Italy). The software automatically extracted gait parameters for each condition, and the faster trial was retained for analysis.

In the DT conditions, participants were instructed on the subtraction rule and given a randomised starting number, limited to a two-digit value greater than 50 (chosen from 51 to 99), before each trial. This restriction was designed to prevent participants from reaching a value of 0 during the counting sequence while walking. Participants counted aloud throughout the walk, without prioritising either the counting or the walking task, and the operator recorded the total responses and errors for each trial.

At the end of the two executions of the same condition (i.e., immediately after completing both trials of ST, DT-1, DT-3, or DT-7), participants provided two domain-specific ratings, one for physical fatigue (VAS P) and one for mental fatigue (VAS M). Each rating was collected using a 20-point visual analogue scale consisting of 20 adjacent boxes (with the first box indicating no fatigue and the rightmost box indicating maximum fatigue). Using a standardised script, participants were instructed to mark a single box that best represented how fatigued they felt for that condition. Ratings were obtained during the seated rest period that followed the second trial of each condition and before instructions for the subsequent condition. For each condition, the chosen box number was recorded as the VAS score (1-20) for VAS P and VAS M, yielding one score per domain per condition. VAS outcomes were analysed across conditions within the same repeated-measures framework used for spatiotemporal gait variables.

3.2.5 Calculations

The impact of dual-tasking on gait performance was demonstrated in two ways.
The first one was via DTC: For each spatiotemporal parameter, the relative cost of performing the DT compared to the single-task condition was calculated using the formula:

$$DTC = \frac{-(DT - ST)}{ST} \times 100$$

A minus sign was placed before the numerator in the subtraction to aid data interpretation. This convention ensures that positive values indicate higher costs during DT compared to ST walking, signalling gait deterioration caused by the cognitive task (Li et al., 2018).

The second way was via Difficulty Index (DI): We developed an additional variable to measure cognitive difficulty. We named this variable the DI, and calculated it as the reciprocal of the number of correct responses per second for each DT condition, using the following formula:

$$DI = \frac{1}{\frac{Number\ of\ response - number\ of\ errors}{Time\ (s) over\ 10m}}$$

Under cognitive load, the number of correct responses within one second will likely decrease, and as a result, the DI will increase, reflecting the relative cognitive challenge of the subtraction tasks paired with walking.

3.2.6 Statistical Analysis

The sample size was determined through an a priori power analysis to estimate the number of participants required to detect changes in gait across different walking conditions. Power analysis was performed using G*Power 3.1 (test family: F tests; statistical test: repeated measures ANOVA, within factors). We assumed a medium effect size (Cohen's $f = 0.25$), $\alpha = .05$, desired power ($1-\beta$) = .80, one group with 4 repeated measurements (ST, DT-1, DT-3, DT-7), a correlation among repeated measures of $r = .50$, and a non-sphericity correction $\varepsilon = 1.00$. Under these assumptions, the analysis indicated that a minimum total sample size of approximately $N \approx 24$ would be sufficient to detect a medium-sized main effect of Condition.

The final sample (N = 30 healthy older adults) exceeded this a priori requirement, thereby meeting the planned ≥80% power to detect a medium effect under the specified assumptions.

All statistical analyses were performed using Jamovi software (version 2.3.2; Sydney, Australia; https://www.jamovi.org). Data were presented as mean ± standard deviation (SD). Demographic characteristics were summarised using descriptive statistics. Differences between sexes were examined with independent-samples t-tests.

A repeated-measures analysis of variance (ANOVA) was conducted to compare spatiotemporal gait parameters across the four walking conditions (ST, DT-1, DT-3, DT-7). When significant main effects were identified, post hoc pairwise comparisons with correction for multiple testing were carried out to determine differences between conditions.

Additionally, linear regression analyses were performed to evaluate the associations between spatiotemporal gait parameters and the calculated DI. Specifically, the slope of these relationships was calculated, and a correlation matrix was created.

Finally, the correlation between DI and mental or physical perception of effort was analysed using correlation analysis.

3.3 Results

Participants' general features:

All participants completed the walking tasks across all modalities (ST, DT-1, DT-3, DT-7) without reporting adverse events. No significant differences in demographic or clinical features were detected between men and women ($p > 0.05$; see *Table 1*).

Spatiotemporal gait parameters:

Table 2 summarises gait parameters across the four walking conditions. Repeated-measures ANOVA revealed significant main effects of test modality on speed, step length, percentage of double support during the gait cycle, and cadence ($p < 0.05$). Specifically, speed, cadence, and step length decreased significantly with task difficulty, while double support increased. Post hoc comparisons confirmed that differences were particularly evident between ST and DT-7 ($p < 0.05$).

Primary Gait Outcomes: Walking speed declined markedly as cognitive load increased ($p < 0.001$), with the most notable reduction observed in DT-7, which differed significantly from all other conditions. Step length followed a similar pattern ($p < 0.001$), gradually shortening from ST to DT-7. Conversely, the percentage of double support increased significantly across conditions ($p < 0.001$), indicating a compensatory strategy to maintain stability under greater cognitive demands. Secondary Gait Outcomes: Cadence also decreased significantly with increasing task difficulty ($p < 0.001$), reflecting a gradual slowdown of step rhythm. No significant main effect was found for step-length CV ($p = 0.088$), although a slight increase was noted in DT-3 and DT-7. Finally, step asymmetry did not differ significantly between conditions ($p = 0.261$), suggesting that task difficulty mainly affected overall timing and stability of gait rather than lateral balance.

	ST	**DT -1**	**DT -3**	**DT -7**	**p**
Speed [m/s]	1.46 ± 0.19	1.38 ± 0.23*	1.27 ± 0.31*#	1.18 ± 0.25*#+	< 0.001
Step length [cm]	73.76 ± 8.64	72.47 ± 9.03*	70.12 ± 9.64*#	66.87 ± 9.99*#+	< 0.001
Double Sup. % GC [%]	26.04 ± 3.8	26.62 ± 3.68	27.82 ± 4.34*#	29.04 ± 4.67*#+	< 0.001
CV Step length [%]	3.45 ± 1.26	3.34 ± 1.24	4.01 ± 1.40	4.16 ± 1.56*#+	0.088
Asymmetry. SX-DX [%]	0.65 ± 4.06	0.36 ± 4.18	-0.30 ± 5.11	-0.16 ± 4.67*#+	0.261
Cadence [Step/min]	119.77 ± 10.96	114.75 ± 13.86	106.01 ± 18.99*#	106.01 ± 13.84*#+	< 0.001

Table 2. Overview of the gait parameter ($p < 0.05$ vs ST; # $p < 0.05$ vs DT-1; + $p < 0.05$ vs DT-3).*

Figure 1 illustrates the DTC for gait parameters across increasing levels of cognitive load.

DTC increased significantly across most gait parameters as task difficulty grew. Primary Gait Outcomes: Speed cost rose considerably ($p < 0.001$), from about 6% in DT-1 to over 20% in DT-7, with post hoc analysis confirming notable differences between DT-1 and both DT-3 and DT-7. Step length cost showed a similar steady rise ($p < 0.001$), reaching its peak in DT-7, with significant differences observed between DT-1 and DT-3 and between DT-3 and DT-7. Double support cost also exhibited a clear upward trend ($p < 0.001$), more than doubling from DT-1 to DT-7, indicating an increasing reliance on stability mechanisms under higher cognitive load. Secondary Gait Outcomes: Cadence cost also rose significantly ($p < 0.001$), with the most notable change between DT-1 and DT-3, and a further increase at DT-7. In contrast, step-length CV cost did not reach statistical significance ($p = 0.103$), although a visible increase was observed at DT-3 and DT-7, suggesting a trend towards greater irregularity in step timing. Finally, asymmetry cost showed no significant effect ($p = 0.407$), and its pattern across conditions was inconsistent, indicating that step asymmetry was not systematically affected by cognitive difficulty.

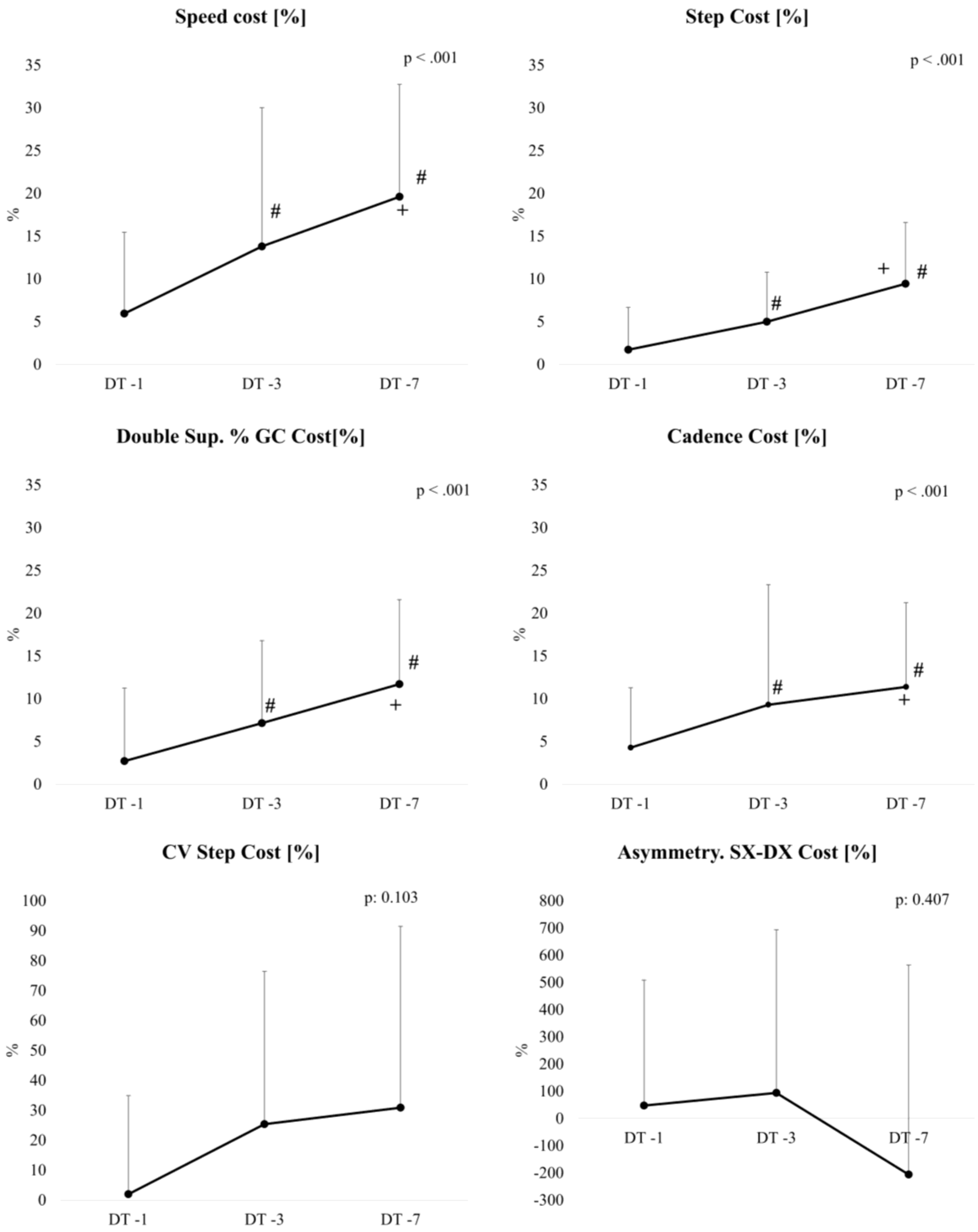

Figure 1. DTC for spatiotemporal gait parameters across conditions (DT-1, DT-3, DT-7). The figure displays mean values ± SD for speed cost, step length cost, double support cost, cadence cost, step-length CV cost, and asymmetry cost. Significant differences are marked (# p < 0.05 vs DT-1; + p < 0.05 vs DT-3).

Table 3 summarises the number of participants falling below the threshold of 1.0 m·s^{-1}, the values of the DI, the number of answers and errors during the subtraction tasks, and the self-reported ratings of physical (VAS P) and mental fatigue (VAS M).

Regarding gait speed, no participant walked below 1.0 m·s^{-1} in the ST condition, while 1 participant did so in DT-1, 7 participants did so in DT-3, and 8 participants did so in DT-7. This confirms that adding a cognitively demanding task progressively impaired safe walking speed, with most subjects approaching or falling below the clinically relevant threshold in the most challenging conditions(Adam et al., 2023; Hainline et al., 2024; Studenski et al., 2011).

The DI increased markedly across DT conditions, with mean values rising from 0.66 ± 0.20 in DT-1 to 1.49 ± 0.70 in DT-3 and 3.45 ± 2.85 in DT-7 ($p < 0.001$). This gradual progression reflects the systematic rise in cognitive demand associated with the subtraction tasks.

Performance on the cognitive task also declined as the difficulty increased. The number of correct answers dropped significantly, from 12.0 ± 2.97 in DT-1 to 6.60 ± 1.90 in DT-3 and 3.97 ± 1.65 in DT-7 ($p < 0.001$). Conversely, errors increased across conditions, reaching the highest value in DT-7 (0.37 ± 0.56; $p = 0.009$).

Self-perceived fatigue scores showed clear patterns for both physical and mental components. VAS P values remained consistent across conditions (ST: 1.4 ± 0.6; DT-7: 1.87 ± 4.17; $p = 0.256$), indicating that the cognitive load did not significantly impact the physical effort of walking. Conversely, VAS M scores demonstrated a notable rise in perceived mental fatigue, increasing from 1.7 ± 1.2 in ST to 4.17 ± 3.90 in DT-7 ($p < 0.001$). Together, these findings suggest that increasing cognitive task difficulty produces a dual effect: it compromises motor performance by reducing gait speed, impacts cognitive performance, and elevates perceived mental fatigue, while largely leaving physical fatigue unaffected.

	ST	**DT -1**	**DT -3**	**DT -7**	**p**
<1.0 [m/s]	0	1	7	8	
Time 10m [s]	6.96 ± 0.95	7.48 ± 1.36*	8.56 ± 3.05*	8.91 ± 2.07$^{*\#}$	<0.001
Answers	-	12.0 ± 2.97	6.60 ± 1.90$^{\#}$	3.97 ± 1.65	<0.001
Errors	-	0.03 ± 0.18	0.23 ± 0.43	0.37 ± 0.56$^{\#}$	0.009
DI	-	0.66 ± 0.20	1.49 ± 0.70$^{\#}$	3.45 ± 2.85$^{\#+}$	<0.001
VAS P	1.4 ± 0.6	1.70 ± 1.73	1.73 ±1.53	1.87 ± 4.17	0.256
VAS M	1.7 ± 1.2	2.43 ± 2.53	3.10 ± 2.77$^{*\#}$	4.17 ± 3.90$^{*\#+}$	<0.001

Table 3. Gait safety threshold (<1.0 m·s^{-1}), DI, cognitive task outcomes (answers and errors), and VAS P and VAS M in the four walking conditions ($p < 0.05$ vs ST; # $p < 0.05$ vs DT-1; + $p < 0.05$ vs DT-3).*

Figure 2 (correlations between the DI and spatiotemporal gait parameters in a representative subject). As task difficulty increased, walking speed demonstrated a strong negative correlation with DI ($r = -0.79 \pm 0.29$; slope = -0.10 ± 0.08 $m \cdot s^{-1}$ per DI unit; intercept $\approx 1.44 \pm 0.22$ $m \cdot s^{-1}$), indicating a gradual decline in velocity. Step length showed a similar trend ($r = -0.77 \pm 0.36$; slope = -2.54 ± 2.11 cm per DI; intercept $\approx 73.75 \pm 8.98$ cm), confirming that higher cognitive loads were associated with shorter steps. Conversely, double support was positively correlated with DI ($r = 0.68 \pm 0.50$; slope = $+1.12 \pm 1.17$ %GC per DI; intercept $\approx 26.09 \pm 3.75$ %GC), aligning with a compensatory shift toward more stable gait as cognitive demands increased. Cadence also correlated negatively with DI ($r = -0.72 \pm 0.32$; slope = -4.60 ± 4.63 $steps \cdot min^{-1}$ per DI; intercept $\approx 117.93 \pm 12.71$ $steps \cdot min^{-1}$), indicating a systematic slowing of step rhythm under increased cognitive load. In comparison, step asymmetry showed no significant relationship with DI ($r = -0.18 \pm 0.52$; slope = -0.44 ± 2.47 %; intercept $\approx 0.57 \pm 3.33$ %), with considerable variability across conditions. Additionally, step-length CV did not display a consistent trend ($r = 0.28 \pm 0.58$; slope = $+0.33 \pm 0.78$ %; intercept $\approx 3.37 \pm 1.23$ %).

Taken together, these regression analyses show that increasing cognitive task difficulty is strongly linked to decreases in velocity, cadence, and step length, as well as increases in double support. In contrast, asymmetry and step-length variability are not consistently affected. Across all participants, this pattern reinforces the connection between cognitive load and the decline in key gait parameters. For each parameter, group-level mean slopes, intercepts, and correlations are summarised in the accompanying *Table 4*.

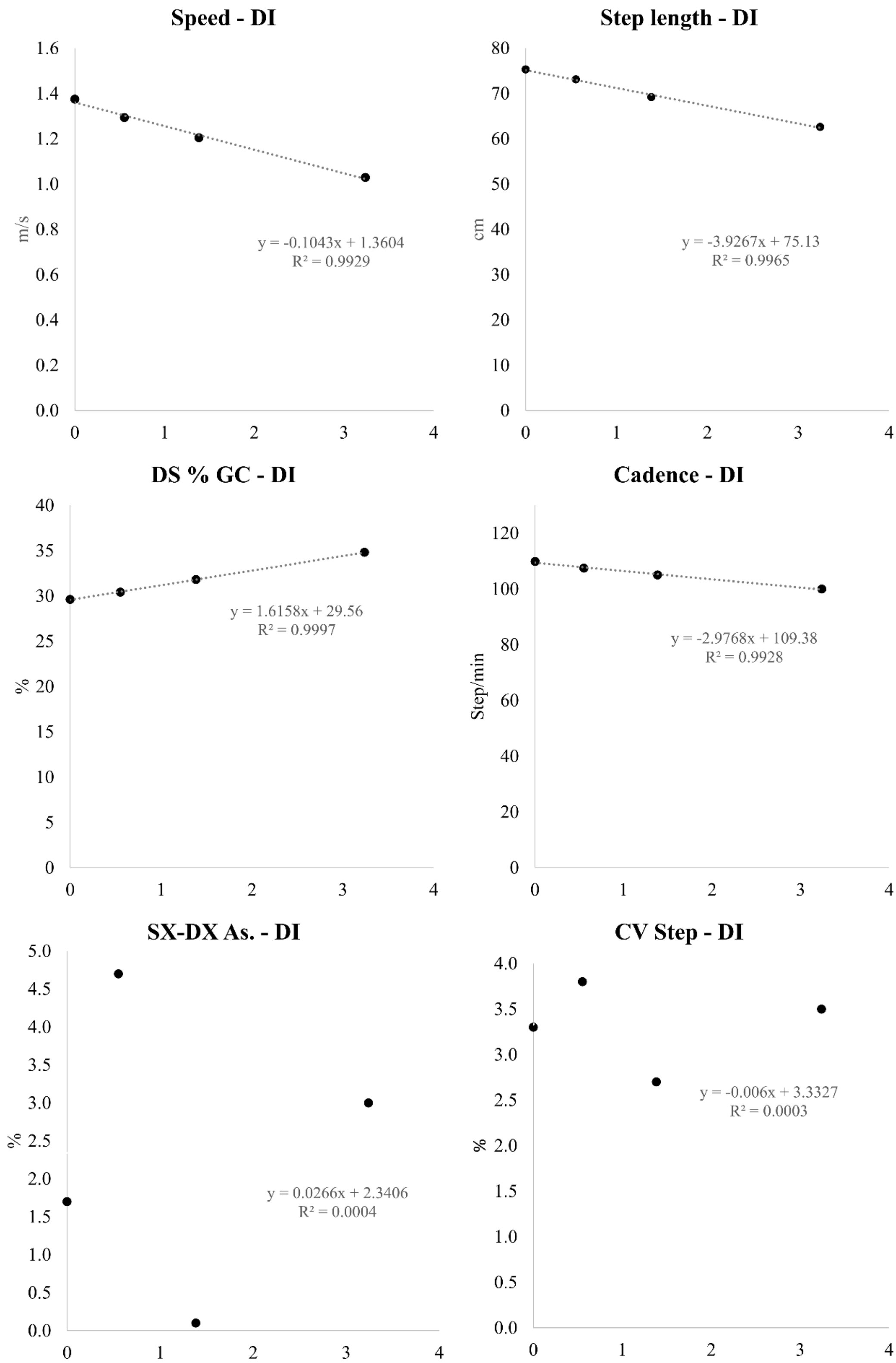

Figure 2. Correlations between a representative subject's DI and gait parameters. The scatterplots show regression lines between DI and speed, step length, double support, cadence, step asymmetry, and step-length CV in absolute units.

	Slope	Intercept	Correlation
Speed [m/s]	-0.10 ± 0.08	1.44 ± 0.22	-0.79 ± 0.29
Step length [cm]	-2.54 ± 2.11	73.75 ± 8.98	-0.77 ± 0.36
Double Sup. % GC [%]	1.12 ± 1.17	26.09 ± 3.75	0.68 ± 0.50
Cadence [Step/min]	-4.60 ± 4.63	117.93 ± 12.71	-0.72 ± 0.32
CV Step length [%]	0.33 ± 0.78	3.37 ± 1.23	0.28 ± 0.58
Asymmetry. SX-DX [%]	-0.44 ± 2.47	0.57 ± 3.33	-0.18 ± 0.52

Table 4. Regression parameters between the DI and gait outcomes.
Slopes, intercepts, and correlation coefficients (mean ± SD) for speed, step length, double support (% gait cycle), cadence, step-length CV, and left-right asymmetry. Slope indicates change in the outcome per unit increase in DI; intercept is the expected value at DI = 0 (≈ single-task).

3.4 Discussion

In this study, we examined how varying levels of cognitive load affect gait in healthy older adults using a standardised DT walking paradigm. Based on capacity-sharing theories of attention, our primary aims were twofold: first, to observe how increasing cognitive demands impair spatiotemporal gait; second, to test two related hypotheses: that DTC varies across difficulty levels (DT-1, DT-3, DT-7), and that within the same individual, the cost increases roughly linearly with task difficulty, reflecting a dose-response relationship between cognitive load and gait control. The main results show that primary spatiotemporal parameters decline gradually as the cognitive task becomes more challenging, with notable effects on velocity, cadence, step length, and double support.

Our results confirm evidence that DT performance causes measurable changes in gait among older adults (Goh et al., 2021; Salzman et al., 2025; Verghese et al., 2007). Specifically, the primary outcomes of velocity and step length are two well-established indicators of gait efficiency that declined with increasing subtraction difficulty. In contrast, double support and step length variability increased, reflecting compensatory strategies to maintain stability. (Montero-Odasso et al., 2012; Yogev-Seligmann et al., 2008). These patterns support the idea that attentional and executive resources are shared across motor and cognitive areas, resulting in interference when cognitive demands exceed available capacity (Al-Yahya et al., 2011; Yogev-Seligmann et al., 2008). This gradual deterioration from ST through DT-7 supports the first hypothesis by showing that higher subtraction loads cause increasingly greater interference, resulting in larger DTC at higher difficulty. The analysis of DTC reinforces this interpretation: relative costs increased alongside task difficulty, supporting the sensitivity of cost-based measures for detecting subtle motor-cognitive interference (Dubost et al., 2006).

Building on this, our DI offers a broad measure of cognitive load that rises with subtraction difficulty; its strong links to velocity, cadence, step length, and double support highlight a clear dose-response relationship between cognitive demand and gait deterioration (Al-Yahya et al., 2011; Goh et al., 2021). By representing DT interference as a continuous measure instead of categorical task categories, DI helps to address the variability in protocols highlighted in previous research and improves comparability across studies (Goh et al., 2021; Plummer et al., 2015). Notably, DI also predicted DTC across parameters, suggesting that a single quantitative marker can summarise the motor-cognitive link more effectively than considering raw conditions alone. These observations align with recent index-based approaches that quantify motor-cognitive load during DT walking, including attentional/automaticity index (Liu et al., 2023), composite performance index developed for DT mobility tests (Brauner et al., 2021), and inertial-based "normalcy" index derived from DTC (Meng et al., 2025). Converging evidence also indicates that empirically ranked increases in cognitive task difficulty are linked with progressively worse gait and postural outcomes (Almutairi et al., 2025). Crucially, and in line with our second hypothesis, individual-level regressions using DI produced positive slopes for the primary gait parameters (lower speed, shorter steps, reduced cadence, and greater double support at higher DI), indicating an approximately linear within-subject dose-response rather than a simple between-condition contrast at the group level.

From a clinical perspective, it is notable that a significant number of participants fell below the 1.0 $m \cdot s^{-1}$ threshold in the more challenging conditions, as walking speeds under 1.0 $m \cdot s^{-1}$ are linked with disability and fall risk in older adults (Verghese et al., 2007; Yogev-Seligmann et al., 2008). The hypothesised dose-response is clinically significant because it helps estimate the additional cognitive load that could push a person towards safety thresholds (e.g., ~1.0 $m \cdot s^{-1}$). Including DI in clinical DT assessments could aid personalised risk assessment (e.g., the chance of falling below 1.0 $m \cdot s^{-1}$ under cognitive strain) and establish a standardised target for tracking changes over time or following rehabilitation (Plummer et al., 2015; Salzman et al., 2025; Verghese et al., 2007).

Together, these results reinforce the idea that DT paradigms provide a sensitive measure of motor-cognitive integration and could act as early indicators of functional decline (Al-Yahya et al., 2011; Salzman et al., 2025). By systematically adjusting cognitive difficulty and quantifying it with a continuous index, this approach supports ongoing standardisation efforts. It offers a structured, reproducible framework for clinical assessment and research applications (Goh et al., 2021; Plummer et al., 2015). More broadly, confirming both the between-condition divergence of costs and the within-subject linear dose-response advances a mechanistic account of how cognitive load interferes with gait, linking experimental manipulation (task difficulty) to predictable changes in control strategies.

3.5 Conclusions

This study shows that increasing cognitive task difficulty leads to progressively measurable impairments in gait performance among healthy older adults. Spatiotemporal parameters declined stepwise from single- to DT conditions, with DTC rising proportionally with cognitive load. Additionally, a group of participants experienced clinically significant reductions in walking speed. These results directly support our experimental hypotheses: DTC varied across difficulty levels (DT-1 < DT-3 < DT-7), and within individuals, costs approximately increased linearly with increased difficulty, as reflected by the DI.

The strong correlations between DI and gait parameters confirm this approach as a dependable way to measure motor-cognitive interference. These findings support the incorporation of standardised graded DT protocols in clinical assessments and research. The framework improves comparability and individualised risk estimation by basing interpretation on both condition-based costs and a continuous difficulty metric (e.g., proximity to the 1.0 $m \cdot s^{-1}$ threshold under cognitive load). Future research should explore the predictive value of these measures for fall risk, extend their application to clinical populations, and investigate their utility in rehabilitation programmes to bolster motor-cognitive resilience (Mou and Jiang, 2025).

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CHAPTER 4: Graded Dual-Task Walking in Parkinson's Disease and Healthy Older Adults: Differential Gait Responses to Increasing Cognitive Load

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Abstract

Dual-task (DT) walking is a sensitive method for assessing motor-cognitive interference in Parkinson's disease (PD), but standardised, graded protocols with measurable cognitive load are lacking. We tested a back-counting DT task with three difficulty levels in 56 community-dwelling individuals with PD (18 females; 65.88 ± 8.7 years; H & Y 1.9 ± 0.63; disease duration 5.3 ± 3.2 years) and 30 healthy older controls (15 females; 69.83 ± 5.22 years). Participants walked 10 metres over ground at self-selected speed under single-task (ST) and DT-1, DT-3, and DT-7 conditions (counting backwards by 1, 3, or 7). Speed, step length, double support time, cadence, and step-length variability were measured, along with a continuous Difficulty Index (DI; net subtraction rate), dual-task costs (DTC), gait safety ($< 1.0\ m\cdot s^{-1}$), and physical/mental fatigue. Linear mixed-effects models (Group, Condition, Group × Condition; Satterthwaite df, Holm-adjusted contrasts) and supplementary regressions of gait/DTC on DI were used. Condition significantly influenced speed, step length, double support, and cadence (all $p < 0.001$), with worsening from ST to DT- 7; PD individuals had slower gait, shorter steps, and greater double support than controls (speed $p = 0.002$ ES = 0.71; step length $p < 0.001$, ES = 0.76; double support $p < 0.001$, ES = 0.86), with Group × Condition interactions for speed ($p = 0.005$) and step length ($p < 0.001$). DI increased from 0.66 ± 0.20 (DT- 1) to 1.49 ± 0.70 (DT- 3) and 3.45 ± 2.85 (DT- 7, $p < 0.001$), while correct answers decreased from 12.0 ± 2.97 to 3.97 ± 1.65 ($p < 0.001$). The number of participants walking < 1.0 $m\cdot s^{-1}$ rose from 0 vs 1 (controls vs PD) at ST to 8 vs 29 at DT- 7. DI showed strong linear relations with gait measures (e.g., speed $r \approx -0.79$ controls, - 0.84 PD) and DTC, with PD showing higher baseline costs and steeper cost-DI slopes for speed and step length. Perceived mental fatigue increased more in PD (Condition $p < 0.001$; Group $p = 0.02$; Group × Condition $p < 0.001$), while physical fatigue changes were modest. Graded subtraction DT walking with a DI provides feasible, clinically relevant dose-response markers of motor-cognitive interference, highlighting poorer baselines and higher costs in PD, while demonstrating largely similar load sensitivity. This may aid standardised assessment, risk stratification, and long-term monitoring.

4.1 Background on Dual-Task Walking Tests

Walking whilst performing a concurrent cognitive task, known as DT walking, acts as a sensitive measure of motor-cognitive interference in ageing and neurological conditions, particularly PD. Under DT conditions, individuals usually exhibit slower gait, shorter steps, and increased double support, while these spatiotemporal changes act as indirect, functional indicators of reduced gait automaticity when attentional resources are stretched (Al-Yahya et al., 2011; Kelly et al., 2012), they primarily reflect the behavioural consequence of motor-cognitive interference rather than its direct neural mechanisms. Recent research demonstrates that DT paradigms also detect more subtle kinematic changes, such as decreased gait smoothness and impaired turning control in PD, thereby strengthening their clinical sensitivity (Caronni et al., 2025). Meta-analytic evidence confirms systematic DT reductions in PD compared to single-task (ST) walking (Raffegeau et al., 2019).

4.1.1 Current Research Gaps

Despite their widespread use, DT protocols differ significantly in cognitive task type and difficulty (e.g., verbal fluency versus serial subtraction), which hampers comparability across studies (Al-Yahya et al., 2011; Tsang et al., 2022). Even within serial subtraction, individual differences in numeracy and baseline cognition can lead to very different subjective loads: tasks that are too easy may result in negligible DT cost, whereas tasks that are too difficult may cause failure and floor effects (Srygley et al., 2009; Tsang et al., 2022). Furthermore, although DT costs often increase with difficulty, relatively few studies have implemented graded difficulty and explicitly modelled dose-response patterns side-by-side in PD and controls, or linked gait changes to a continuous measure of cognitive load (Raffegeau et al., 2019; Zhou et al., 2025).

4.1.2 Development of a Dual-Task Walking Test with Three Levels of Difficulty

A practical approach to standardise cognitive load is to evaluate the serial subtraction task at multiple levels (e.g., −1, −3, -7). This facilitates level-wise comparisons and systematic trend analysis while decreasing variability between subjects in perceived load (Srygley et al., 2009). Recent research suggests ranking and calibrating DT cognitive tasks based on perceived difficulty and performance traits, offering empirical guidance for choosing and sequencing DT challenges (Almutairi et al., 2025). Together, these approaches help reduce under- and over-loading, thereby increasing measurement reliability.

4.1.3 Study Rationale and Utility

A graded, back-counting DT battery can address three key questions in PD: whether PD exhibits worse baseline gait under ST; whether increasing cognitive load results in consistent declines in gait (dose-response); and whether PD shows disproportionate sensitivity to load (group × difficulty interaction). Adjusting difficulty levels and, when possible, summarising load with a continuous DI can enhance interpretability and comparability across sessions and cohorts (Kelly et al., 2012; Raffegeau et al., 2019). Converging evidence links executive control with DT gait performance, supporting the use of cognitively graded paradigms and index that reflect attentional demand (Zhou et al., 2025).

4.1.4 Broader Applications

Standardised, graded DT gait tests are useful beyond characterising PD: they can screen for motor-cognitive interference in older adults, monitor long-term changes, and act as outcomes in rehabilitation trials. Recent reviews show that DT focused training enhances DT gait performance in PD (Sarasso et al., 2024; Wong et al., 2023), though some reviews note that advantages over single-task training may depend on dosage, task specifics, and transfer effects (Lin et al., 2024). Adjunct strategies like sensory cueing may further influence performance under DT, though evidence remains mixed (Azoidou et al., 2024). More generally, exercise meta-analyses emphasise that targeted modalities and appropriate dosing enhance ambulatory function in PD, offering a supportive context for DT-oriented assessment and training (Xie et al., 2025; Zhen et al., 2022).

4.1.5 Study Aim and Experimental Hypotheses

We developed a graded DT walking test with three subtraction levels (−1, −3, -7) to assess feasibility and validity in individuals with early-stage PD and age-matched controls. We hypothesised that: under ST, PD would show lower speed and step length and greater double support time than controls; DT costs would increase systematically with task difficulty (dose-response) in both groups; and the slope of the relationship between a continuous DI and gait outcomes would be similar across groups, indicating comparable despite lower baseline performance in PD (Almutairi et al., 2025; Kelly et al., 2012; Raffegeau et al., 2019; Srygley et al., 2009; Zhou et al., 2025).

4.2 Materials and Methods

4.2.1 Participants

We enrolled fifty-six community-dwelling older adults with PD (18 females; 65.88 ± 8.7 years; 1.9 ± 0.63 H & Y) and thirty healthy controls (CG) (15 females; 69.83 ± 5.22 years). Participants were recruited through flyers, local advertisements, and word of mouth; PD participants were referred from outpatient clinics. Inclusion criteria for both groups included: no current or recent (≤ 6 months) orthopaedic injury; no acute neurological events; no lower-limb pain limiting walking; and no cognitive impairment, defined as Mini-Mental State Examination (MMSE) > 24 (Ramaker et al., 2002).

A trained neurologist diagnosed all PD patients according to the UK Parkinson's Disease Society Brain Bank clinical diagnostic criteria (Hughes et al., 1992) and classified them as being in a stage below 3.0 of the Hoehn and Yahr scale (H & Y; Hoehn and Yahr, 1967). Disease duration was recorded (5.3 ± 3.2 years), and at the time of testing, all patients were in the on-phase of anti-Parkinsonian medication. There was no fixed interval for testing relative to the last medication intake because the participants varied in the type of dopaminergic medication used and, consequently, in the medications' duration of action, as is known from other research (Salazar et al., 2017).

Mobility limitation was evaluated using the Short Physical Performance Battery (SPPB) (Guralnik et al., 1994). Health-related quality of life was evaluated using the SF-36, with standardised Physical (PCS) and Mental (MCS) Component Summary scores (Apolone and Mosconi, 1998; Garratt and Stavem, 2017). Psychological screening involved the State-Trait Anxiety Inventory (STAI-Y; trait and state forms) and the Beck Depression Inventory-II (BDI-II) (Bieling et al., 1998; Linde et al., 2022; Wang and Gorenstein, 2013). Sleep quality was measured with the Pittsburgh Sleep Quality Index (PSQI).

After screening, 30 healthy older adults (15 females) and 56 individuals with PD (18 females) were included. Group summaries are reported in *Table 1*. Briefly, groups were broadly comparable in anthropometrics (weight, height, BMI) and global cognition (MMSE), with PD showing younger age, fewer years of education, higher depressive symptoms (BDI-II), poorer sleep quality (PSQI), and lower SF-36 PCS/MCS than controls; other questionnaire scores were similar across groups. PD clinical characteristics are also reported (disease duration, Hoehn and Yahr stage, MDS-UPDRS part III).

Mean ± SD	**CG**	**PD**	***p***
N° (Female)	30 (15)	56 (18)	-
Age [yr]	69.83 ± 5.22	65.88 ± 8.7	**0.03**
Weight [kg]	72.33 ± 13.47	74.13 ± 15.21	0.59
Height [cm]	167.96 ± 9.91	169.26 ± 9.85	0.56
BMI [$kg \cdot m^{-2}$]	25.58 ± 4.1	25.74 ± 4.71	0.88
Years of Education [yr]	13.43 ± 2.7	11.94 ± 3.8	0.06
BDI	3.63 ± 4.54	8.7 ± 8.05	**<0,001**
STAY_1	30.7 ± 5.25	34.25 ± 8.71	0.07
STAY_2	37.03 ± 9.07	37.25 ± 10.19	0.92
MMSE	27.1 ± 1.34	27.09 ± 1.88	0.97
PSQI	4.97 ± 2.54	6.4 ± 3.55	0.05
SPPB	12 ± 0	11.65 ± 0.91	**0.04**
SF-36 PCS	52.54 ± 6.22	46.71 ± 6.7	**<0,001**
SF-36 MCS	52.47 ± 5.45	45.43 ± 8.55	**<0,001**
PD duration [yr]		5.09 ± 3.83	
H&Y		1.9 ± 0.63	
UPDRS-3		28.7 ± 10.84	

Table 1. Participant Characteristics. Group means ± SD for demographics, clinical and questionnaire measures in healthy controls (CG) and Parkinson's disease (PD), with between-group p-values. PD clinical severity index (duration, H & Y, MDS-UPDRS III) are reported descriptively.

4.2.2 Ethical Approval

The study was conducted in accordance with the Declaration of Helsinki and approved by the University of Verona Committee for the Approval of Research on Humans (CARU; protocol no. 08/2022). All participants provided written informed consent before taking part.

4.2.3 Study Design and Procedure

Participants attended two study sessions conducted by trained investigators at a similar time of day (10:00 a.m. ± 2 hours) in an environmentally controlled exercise physiology laboratory at the University of Verona (temperature: 22-25°C, relative humidity: 55-65%). A 48-hour recovery period was required between the sessions. To minimise any time-related changes in participants' physiology, all visits were completed within 7 days. Demographical and clinical features (anthropometrics, cognition, functional independence, perceived health, psychological symptoms, and sleep quality) were collected during the initial session, while in the second session, participants performed overground 10-m walking trials at their self-selected comfortable speed under: ST walking and DT back-counting at three difficulty levels (−1, −3, −7), presented in randomised order. Standardised instructions emphasised maintaining the usual walking speed while performing the cognitive task aloud. Rest periods were provided as needed to prevent fatigue.

4.2.4 Anamnestic Session: Demographic and Clinical Data

A structured interview and validated questionnaires were utilised to characterise the sample. Anthropometric measures included height (168.0 ± 9.9 cm, measured using a vertical stadiometer, Seca, Leicester, UK) and weight (72.1 ± 13.2 kg, measured using a digital scale, Seca, Leicester, UK). Data on handedness and years of education were collected.

4.2.5 Experimental Session: Dual-Task Walking Protocol

Participants performed four different walking conditions in randomised order:

- Single-task (ST): walking without a concurrent task.
- Dual-task 1 (DT-1): walking while counting backwards by ones.
- Dual-task 3 (DT-3): walking while counting backwards by threes.
- Dual-task 7 (DT-7): walking while counting backwards by sevens.

Each condition was performed twice, resulting in eight trials per participant, with one-minute seated rests between trials. Walking trials took place on a 10-metre walkway. To ensure measurement of steady-state gait, participants began walking 1.5 metres before entering the sensor area and continued for 1.5 metres beyond its end. Participants were instructed to walk at a self-selected pace along a straight path.

Spatiotemporal gait parameters, including time to cover 10 metres, speed ($m \cdot s^{-1}$), cadence (steps/min), step length (cm), double support time (% gait cycle), and step length coefficient of variation (step-length CV), were measured using a validated photoelectric system (Witty Gate, Microgate, Bolzano, Italy) integrated with 10 one-metre photoelectric bars and dedicated software (OptoGait, Microgate, Bolzano, Italy). The software automatically extracted gait parameters for each condition, and the faster trial was retained for analysis.

In the DT conditions, participants were instructed on the subtraction rule and given a randomised starting number, limited to a two-digit value greater than 50 (drawn from 51-99), before each trial. This limitation was used to prevent participants from reaching zero during the walking part of the counting sequence. Participants counted aloud throughout the walk, without prioritising either the counting or walking task, and the operator recorded the total responses and errors for each trial.

At the end of the two executions of the same condition (i.e., immediately after completing both trials of ST, DT-1, DT-3, or DT-7), participants provided two domain-specific ratings, one for VAS P and one for VAS M. Each rating was collected using a 20-point visual analogue scale consisting of 20 adjacent boxes (with the first box indicating no fatigue and the rightmost box indicating maximal fatigue). Using a standardised script, participants were instructed to mark a single box that best represented how fatigued they felt for that condition. Ratings were gathered during the seated rest period that followed the second trial of each condition and prior to instructions for the subsequent condition. For each condition, the chosen box number was recorded as the VAS score (1-20) for VAS P and VAS M, providing one score per domain per condition. VAS outcomes were analysed across conditions within the same repeated-measures framework used for spatiotemporal gait variables.

4.2.6 Outcome Measures

Spatiotemporal gait parameters included speed, step length, and percentage of double support (%DS). GV and SL were measured using a photoelectric sensor walkway; start and finish times were synchronised with photocells at the 10-m course boundaries. %DS was calculated from contact and flight timings derived from the photoelectric system, based on the manufacturer's algorithms. For DT trials, we calculated:

$$DTC\ (\%) = \frac{-(DT - ST)}{ST} \times 100$$

Sign adapted so that positive values indicate worsening performance, where appropriate.

A continuous Difficulty Index (DI) reflecting the effective cognitive load of the subtraction task (Li et al., 2018):

$$DI = \frac{1}{\frac{Number\ of\ response - number\ of\ errors}{Time\ (s) over\ 10m}}$$

where "subtractions - errors per second" quantifies the net calculation rate.

4.2.7 Statistical Analysis

The sample size was determined through an a priori power analysis to estimate the number of participants required per group, assuming a medium effect size for the Group × Condition interaction. Power analysis was performed using G*Power 3.1 (test family: F tests; statistical test: repeated measures ANOVA, within-between interaction). We specified a medium effect (Cohen's $f = 0.25$), $\alpha = .05$, power $(1-\beta) = .80$, 2 groups (PD, healthy controls) and 4 repeated measurements (ST, DT-1, DT-3, DT-7), assuming a correlation among repeated measures $r = .50$ and $\varepsilon = 1.00$. Under these assumptions, the analysis indicated that a total sample of approximately $N \approx 50$ (about 25 participants per group) would be sufficient to detect a medium-sized Group × Condition interaction. The final sample ($N = 86$; 56 individuals with PD and 30 healthy controls) exceeded this a priori requirement, thereby meeting the planned ≥80% power to detect a medium effect under the specified assumptions. Analyses were performed using linear mixed-effects models (LMMs) with restricted maximum likelihood (REML). For each gait outcome (speed, step length, and percentage of double support), we fitted a model that included fixed effects for Group (PD, CG) and Condition (ST, DT-1, DT-3, DT-7), as well as their interaction (Group × Condition). To account for within-subject correlation, a random intercept for participant (ID) was included. If the model converged and showed an improved fit (ΔAIC), a random slope for Condition was also added. Degrees of freedom and p-values for fixed effects were calculated using the Satterthwaite approximation. When appropriate, estimated marginal means (EMMs) were compared with Holm-adjusted pairwise tests within the relevant family of contrasts (e.g., PD vs CG at each condition; condition-wise comparisons within groups). Model residuals were examined for normality and homoscedasticity; if necessary, outcomes were transformed, or a heterogeneous residual variance by group/condition was specified.

To isolate the cognitive component, we repeated the analysis on DTC with the Condition limited to the three DT levels (DT-1, DT-3, and DT-7) and the Group as before. In a complementary load-response analysis, we modelled the DI as a centred continuous covariate, testing the Group × DI interaction for differential sensitivity to cognitive load; participant random intercepts (± DI slopes) were included similarly. For completeness, we report fixed-effect estimates, standard errors, 95% CIs, and standardised effect sizes (e.g., semi-partial R^2 for mixed models or Cohen's d for simple contrasts). The significance threshold was $\alpha = 0.05$. When multiple outcomes were analysed, inferences were made for each; exploratory cross-outcome comparisons were summarised with the FDR (Benjamini-Hochberg) method when appropriate.

Group comparisons of sample characteristics (*Table 1*) were conducted using t-tests or χ^2 tests as appropriate; these were purely descriptive and not adjusted for multiple comparisons. Sensitivity analyses involved adding age, years of education, BDI-II, PSQI, and SF-36 PCS/MCS as between-subject covariates if imbalances suggested potential confounding; however, the results remained consistent and are available upon request.

4.3 Results

General features of participants: All participants completed the walking tasks across all modalities (ST, DT-1, DT-3, DT-7) without reporting adverse events. Linear mixed-effects models indicated a significant effect of Condition with the fixed factors explaining a substantial proportion of variance (marginal R^2, R_m^2) for speed (R_m^2 = 0.227), step length (R_m^2 = 0.206), double support (R_m^2 = 0.200), and cadence (R_m^2 = 0.179) (all $p < 0.001$). Meanwhile, step-length CV did not differ across conditions (p = .167). Throughout the four tests, speed, step length, and cadence decreased gradually from ST to DT-7, while double support increased with task difficulty.

There was a clear between-group pattern: compared with CG, participants with PD walked more slowly, with shorter steps and greater double support throughout the protocol. The unstandardised effect sizes (mean differences) and Cohen's d highlight the clinical magnitude of this impairment: PD individuals exhibited a mean speed reduction of -0.16 m/s (p = 0.002, ES = 0.71), a step length reduction of -6.93 cm ($p < 0.001$, ES = 0.76), and an increase in double support of +3.52 %GC ($p < 0.001$, ES = 0.86). Cadence showed only a trend towards lower values in PD (p = 0.078). Step-length CV was significantly higher in PD than in CG (main effect of Group $p < 0.001$), with no modulation by condition.

Critically, group × Condition interactions were significant for speed (p = 0.005) and step length ($p < 0.001$), indicating disproportionate declines in PD as cognitive load increased. No interaction was found for double support (p = 0.240), cadence (p = 0.147), or step-length CV (p = 0.880), suggesting similar load-related patterns in both groups for these outcomes.

Post-hoc EMM comparisons (Holm-corrected) mirrored the symbol key in *Table 2*: differences were most evident under DT-7, which differed from all other conditions within each group for speed, step length and cadence, and showed the most significant increase in double support. Between-group contrasts (CG vs PD) became apparent as early as DT-1. They persisted through DT-3/DT-7 for the main spatiotemporal variables (indicated by symbols "$" in *Table 2*), whereas step-length CV was consistently higher in PD across all DT conditions. Overall, these findings confirm a dose-response deterioration of gait with increasing cognitive load, with PD exhibiting lower baseline performance and greater sensitivity to the key spatial-temporal metrics.

	ST		DT -1		DT -3		DT -7		Test	Gr.	X
	CG	PD	CG	PD	CG	PD	CG	PD			
Speed [m/s]	1.46± 0.19	1.37 ± 0.19	1.38 ± 0.23*	1.16 ± 0.25$^{*\$}$	1.27 ± 0.31$^{*\#}$	1.09 ± 0.25$^{*\#\$}$	1.18 ± 0.25$^{*\#+}$	1.01 ± 0.23$^{*\#+\$}$	<.001	0.002	0.005
Step length [cm]	73.76 ± 8.64	69.33 ± 8.44	72.47 ± 9.03	64.42 ± 9.65$^{*\$}$	70.12 ± 9.64$^{*\#}$	62.3 ± 9.3$^{*\#\$}$	66.87 ± 9.99$^{*\#+}$	59.45 ± 8.99$^{*\#+\$}$	<.001	<.001	<.001
Double Sup. % GC [%]	26.04 ± 3.83	28.73 ± 3.94	26.62 ± 3.68	30.52 ± 4.69$^{*\$}$	27.82 ± 4.34*	31.68 ± 4.6$^{*\#\$}$	29.04±4.67$^{*\#}$	32.66 ± 4.42$^{*\#\$}$	<.001	<.001	0.240
Cadence [Step/min]	119.77 ± 10.96	117.4 ± 7.79	114.75 ± 13.86	107.18 ± 13.5*	108.64 ± 18.99*	103.58 ± 15.79$^{*\#}$	106.01 ± 13.84$^{*\#}$	101.30 ± 3.67$^{*\#}$	<.001	0.078	0.147
CV Step length [%]	3.45 ± 1.26	8.52 ± 7.78$^{\$}$	3.34 ± 1.24	9.15 ± 7.36$^{\$}$	4.01 ± 1.4	9.89 ± 8.06$^{\$}$	4.16±1.56	9.94 ± 6.38$^{\$}$	0.167	<.001	0.880

*Table 2. Spatiotemporal gait parameters (mean ± SD) in controls (CG) and Parkinson's disease (PD) across ST, DT-1, DT-3, and DT-7. P-values from a linear mixed model Symbols: $ vs CG; * vs ST; # vs DT-1; + vs DT-3.*

Figure 1 shows DTC for gait parameters across increasing cognitive load (DT-1, DT-3, DT-7). Condition effects were significant for speed cost, step-length cost, double-support cost, and cadence cost (all $p < 0.001$), and for step-length CV cost ($p = 0.006$).

Speed cost increased steadily with difficulty (test < 0.001), with post-hoc differences between DT-1 and DT-3, and between DT-1 and DT-7 (symbols #, +). PD exhibited higher DTC than CG overall ($p = 0.004$), while the Group × Condition interaction was not significant ($p = 0.347$), suggesting similar load-response slopes across groups.

Step-length cost followed the same graded pattern (test < 0.001), with significant increases from DT-1 to DT-3 and from DT-3 to DT-7 (#, +). PD showed greater costs than CG (group < 0.001) without interaction (interaction = 0.885). Double-support cost increased with difficulty (test < 0.001), especially between DT-1 vs DT-3 and DT-1 vs DT-7 (#, +). There was no group difference (group = 0.102) or interaction (interaction = 0.759).

Cadence cost also increased across conditions (test < 0.001), with the steepest change observed between DT-1 and DT-3 (#) and a further rise at DT-7 (+). Group and interaction effects were not significant (group = 0.115; interaction = 0.357). For step-length CV cost, the overall effect of the condition was significant (test = 0.006), with a notable rise towards DT-7. The group effect showed a trend (group = 0.056), and there was no significant interaction (interaction = 0.690). Overall, DTCs exhibited a dose-response increase as cognitive demand intensified. PD participants consistently showed higher costs than controls for speed and step length, with similar load sensitivity (non-significant interactions) across all outcomes.

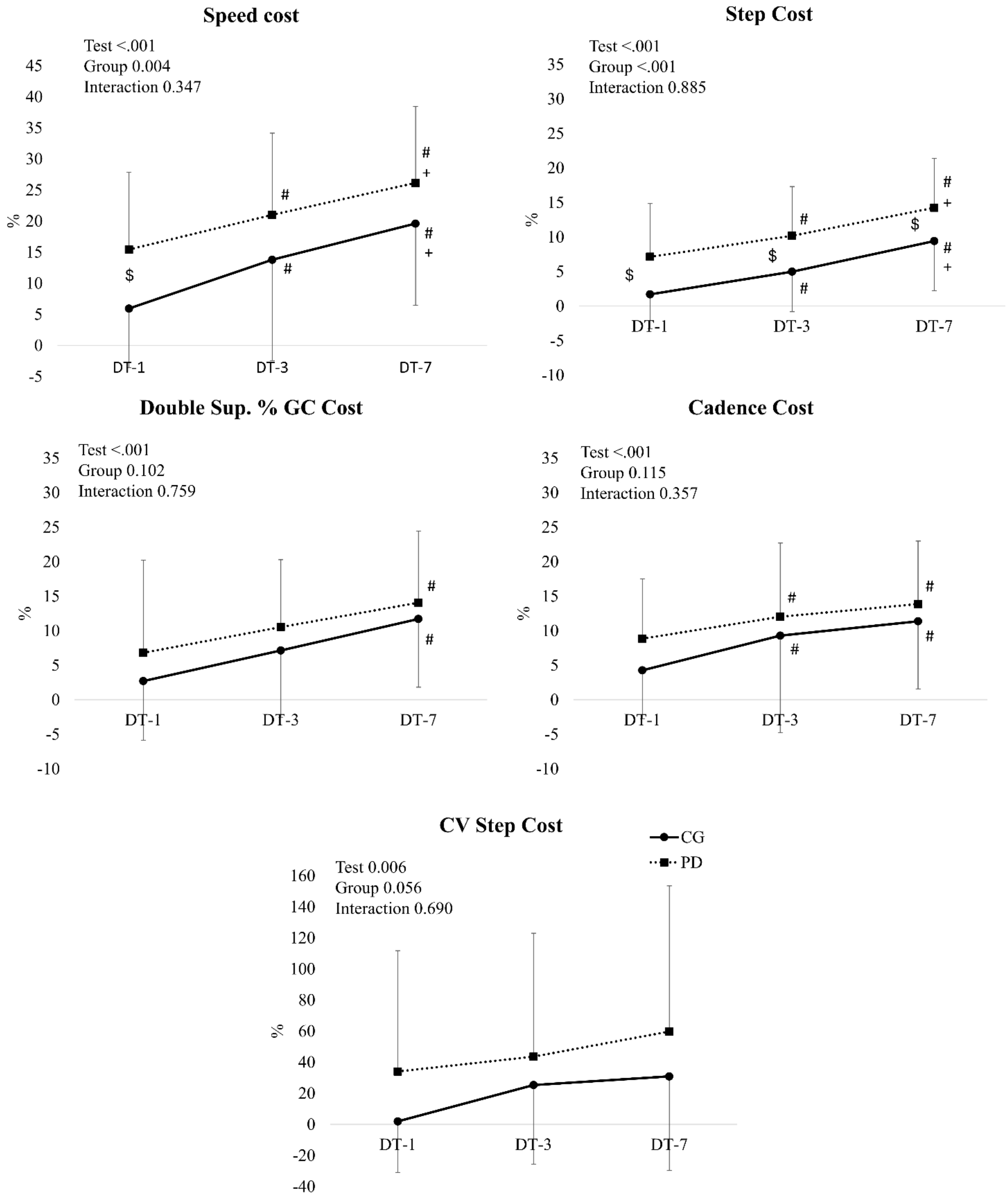

*Figure 1. DTC for spatiotemporal gait parameters across DT-1, DT-3, and DT-7 in controls (CG, circles/solid line) and Parkinson's disease (PD, squares/dotted line). Points = means; error bars = SD. Statistics are from linear mixed models (fixed: Group, Condition, Group×Condition; Satterthwaite df). Symbols: $ vs CG (same condition); # vs DT-1 (same group); + vs DT-3 (same group). post-hoc EMMs Holm-corrected. Symbols: $ vs CG; * vs ST; # vs DT-1; + vs DT-3.*

Table 3 summarises gait safety ($<1.0\ \mathrm{m \cdot s^{-1}}$), cognitive-task performance (answers, errors), DI, and self-reported physical (VAS P) and mental (VAS M) fatigue across DT levels, using linear mixed-model statistics (fixed effects: Group, Condition, Group×Condition; Satterthwaite df).

Gait safety: no control (CG) fell below $1.0\ \mathrm{m \cdot s^{-1}}$ at ST, whereas one PD participant did so at ST. As difficulty increased, the number below $1.0\ \mathrm{m \cdot s^{-1}}$ rose stepwise (DT-1: CG = 1, PD = 11; DT-3: CG = 7, PD = 17; DT-7: CG = 8, PD = 29), indicating a progressive reduction in clinically safe walking speed (Adam et al., 2023; Hainline et al., 2024; Studenski et al., 2011).

Walking time (10 m): the time to cover 10 m increased with difficulty (test $p < 0.001$), was higher in PD compared to CG overall (group $p = 0.010$), and showed a Group × Condition interaction ($p = 0.030$), indicating that the DT-related slowing was more pronounced in PD. Post-hoc contrasts (symbols in *Table 3*) confirmed differences versus ST (*) and between difficulty levels (#, +), with PD already slower than CG in DT-1 ($). Cognitive-task performance: correct answers declined across DT levels (test $p < 0.001$) with no group or interaction effects ($p = 0.07$ and 0.79). Errors increased with difficulty (test $p < 0.001$), with no reliable group difference ($p = 0.19$) and a trend towards a Group×Condition interaction ($p = 0.06$).

DI increased steadily and significantly from DT-1 to DT-7 (test $p < 0.001$), with no differences between groups or interaction effects ($p = 0.98$ each), indicating similar effective cognitive load for PD and CG at each level. Perceived fatigue: VAS P (physical) showed a small but significant effect of condition (test $p = 0.001$), with no difference between groups ($p = 0.83$) and a trend towards interaction ($p = 0.08$); changes were modest across levels. VAS M (mental) rose markedly with difficulty (test $p < 0.001$), was higher in PD overall (group $p = 0.02$), and displayed a Group×Condition interaction ($p < 0.001$), suggesting a steeper increase in perceived mental fatigue in PD (symbols *, #, +).

Overall, increasing cognitive difficulty had a dual effect: a slower and less safe gait (more participants $<1.0\ \mathrm{m \cdot s^{-1}}$; longer 10-m times) and decreased cognitive performance, accompanied by a significant rise in mental fatigue (especially in PD), while physical fatigue also played a part.

	ST		DT -1		DT -3		DT -7		Test	Gr.	X
	CG	PD	CG	PD	CG	PD	CG	PD			
<1.0 m/s	0	1	1	11	7	17	8	29			
Time 10m [s]	6.96 ± 0.95	7.45 ± 1.09	7.48 ± 1.36	$9.12 \pm 2.65^{*\$}$	$8.56 \pm 3.05^{*\#}$	$9.84 \pm 3.02^{*}$	$8.91 \pm 2.07^{*\#}$	$10.56 \pm 3.03^{*\#\$}$	<.001	0.01	0.03
Answers	-	-	12 ± 2.97	12.79 ± 2.99	$6.6 \pm 1.9^{\#}$	$7.71 \pm 3.25^{\#}$	$3.97 \pm 1.65^{\#+}$	$4.73 \pm 1.89^{\#+}$	<.001	0.07	0.79
Errors	-	-	0.03 ± 0.18	0.09 ± 0.35	0.23 ± 0.43	0.18 ± 0.43	0.37 ± 0.56	$0.66 \pm 0.75^{\#+}$	<.001	0.19	0.06
DI			0.66 ± 0.2	0.75 ± 0.24	1.49 ± 0.7	$1.49 \pm 0.73^{\#}$	$3.45 \pm 2.85^{\#+}$	$3.38 \pm 2.18^{\#+}$	<.001	0.98	0.98
VAS P	1.4 ± 0.62	2 ± 1.58	1.7 ± 1.73	2.38 ± 2.19	1.73 ± 1.53	2.57 ± 2.1	1.87 ± 1.74	2.48 ± 2.09	0.00	0.83	0.08
VAS M	1.7 ± 1.21	1.84 ± 1.72	2.43 ± 2.53	$2.95 \pm 2.89^{*}$	3.1 ± 2.77	$4.63 \pm 3.45^{*\#}$	$4.2 \pm 3.9^{*}$	$7.05 \pm 4.56^{*\#+}$	<.001	0.02	0.00

*Table 3. Gait safety threshold (<1.0 m·s⁻¹), 10-m time, Difficulty Index (DI), cognitive-task outcomes (answers, errors), and perceived fatigue (VAS P, VAS M) across ST, DT-1, DT-3, DT-7 in controls (CG) and Parkinson's disease (PD). P-values from a linear mixed model (fixed: Group (Gr), Test, Group×Condition (X); Satterthwaite df); post-hoc EMMs Holm-corrected. Symbols: $ vs CG; * vs ST; # vs DT-1; + vs DT-3.*

Figure 2 (below) illustrates the relationships between the DI and spatiotemporal gait parameters for a typical participant. As cognitive load increases, walking speed decreases, while DI and step length reduce, and double support time increases linearly, aligning with a compensatory shift towards greater stability. Cadence also decreases as DI rises, indicating a systematic slowing of step rhythm. In contrast, step-length CV does not display a consistent DI-related trend. Note that these plots are for one representative participant; group-level mean slopes and intercepts for CG and PD are reported separately in *Table 4*.

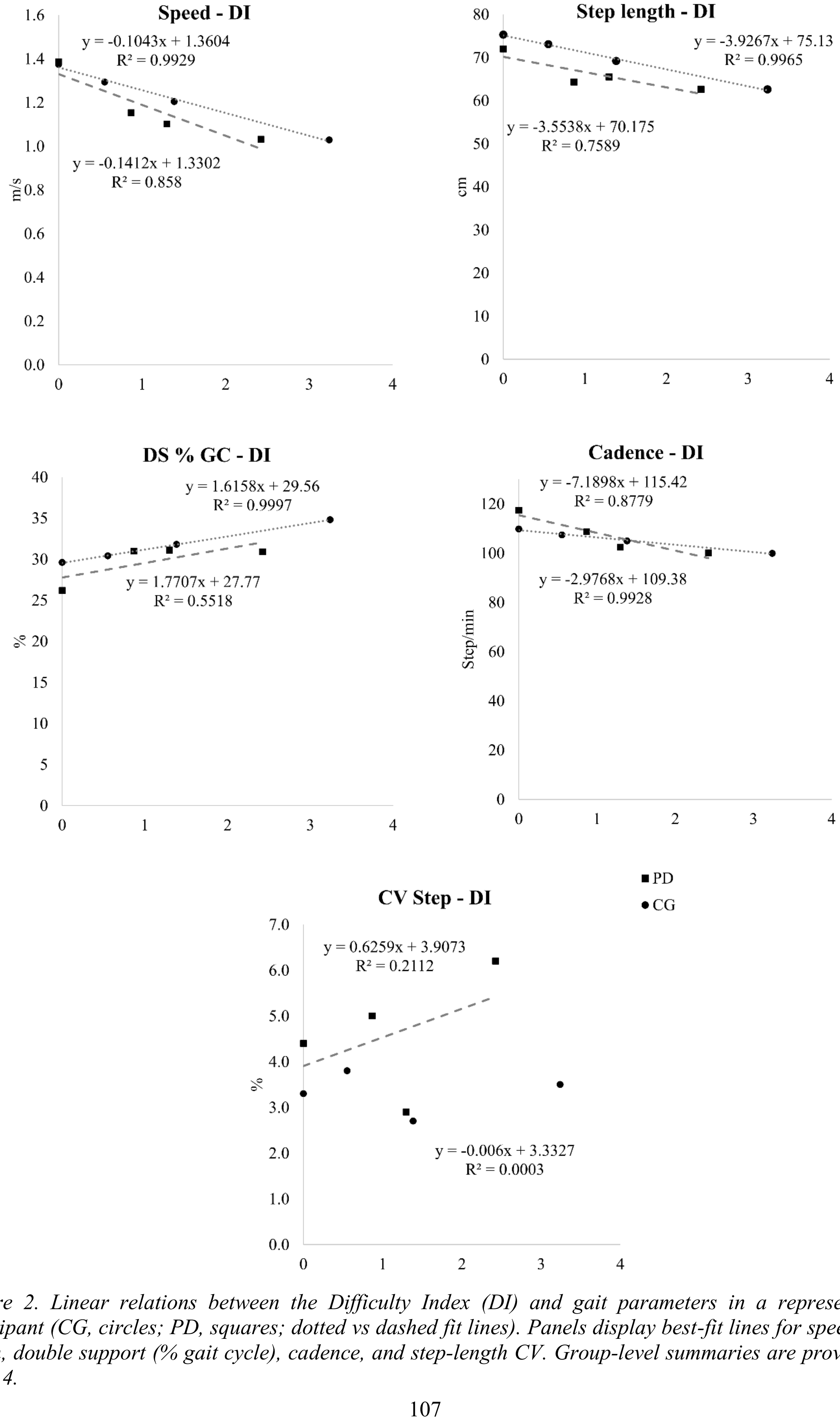

Figure 2. Linear relations between the Difficulty Index (DI) and gait parameters in a representative participant (CG, circles; PD, squares; dotted vs dashed fit lines). Panels display best-fit lines for speed, step length, double support (% gait cycle), cadence, and step-length CV. Group-level summaries are provided in Table 4.

Table 4 presents the group-level parameters derived from the linear fits between DI and gait outcomes. For speed, both groups exhibited a strong negative correlation with DI (CG r = −0.79 ± 0.29; PD r = −0.84 ± 0.19), with comparable slopes (CG −0.10 ± 0.08 $m \cdot s^{-1}$/DI; PD −0.14 ± 0.09 $m \cdot s^{-1}$/DI). However, the intercept was lower in PD (1.31 ± 0.22 $m \cdot s^{-1}$) compared to controls (1.44 ± 0.22 $m \cdot s^{-1}$; $), suggesting a reduced baseline velocity.

Step length followed a similar pattern: correlations were negative and substantial (CG −0.77 ± 0.36; PD −0.85 ± 0.16). Here, PD exhibited a steeper (more negative) slope (−3.85 ± 2.70 cm/DI) than CG (−2.57 ± 2.09 cm/DI; $), along with a lower intercept (68.25 ± 8.87 vs 73.71 ± 9.05 cm; $), consistent with shorter baseline steps and greater load-related shortening in PD.

For double support, correlations were positive in both groups (CG 0.68 ± 0.50; PD 0.69 ± 0.38), and slopes were similar (CG +1.13 ± 1.16; PD +1.58 ± 1.44 %GC/DI). The intercept was higher in PD (29.15 ± 3.82%) than in CG (26.11 ± 3.77%), indicating greater baseline double support. Cadence correlated negatively with DI in both groups (CG −0.70 ± 0.32; PD −0.76 ± 0.31); slopes (−4.59 ± 4.64 vs −5.95 ± 4.67 $steps \cdot min^{-1}$/DI) and intercepts (117.98 ± 12.73 vs 113.71 ± 9.76 $steps \cdot min^{-1}$) did not differ significantly between groups. Finally, step-length CV showed weak, inconsistent associations with DI (CG r = 0.28 ± 0.58; PD r = 0.40 ± 0.56), with similar slopes (0.35 ± 0.78 vs 0.78 ± 2.28 %/DI) but a higher intercept in PD (8.58 ± 7.56%) compared with CG (3.70 ± 2.30%), reflecting elevated baseline variability.

	Slope		**Intercept**		**Correlation**	
	CG	PD	CG	PD	CG	PD
Speed [m/s]	-0.1 ± 0.08	-0.14 ± 0.09	1.44 ± 0.22	1.31 ± 0.22$	-0.79 ± 0.29	-0.84 ± 0.19
Step length [cm]	-2.57 ± 2.09	-3.85 ± 2.7 $	73.71 ± 9.05	68.25 ± 8.87$	-0.77 ± 0.36	-0.85 ± 0.16
Double Sup. % GC [%]	1.13 ± 1.16	1.58 ± 1.44	26.11 ± 3.77	29.15 ± 3.82$	0.68 ± 0.5	0.69 ± 0.38
Cadence [Step/min]	-4.59 ± 4.64	-5.95 ± 4.67	117.98 ± 12.73	113.71 ± 9.76	-0.7 ± 0.32	-0.76 ± 0.31
CV Step length [%]	0.35 ± 0.78	0.78 ± 2.28	3.7 ± 2.3	8.58 ± 7.56$	0.28 ± 0.58	0.4 ± 0.56

Table 4. Group-level mean (± SD) slope, intercept, and correlation (r) from linear fits between the Difficulty Index (DI) and gait variables in controls (CG) and Parkinson's disease (PD). Units Slope: Speed ($m \cdot s^{-1}$ per DI), Step length (cm per DI), Double support (%GC per DI), Cadence ($steps \cdot min^{-1}$ per DI), step-length CV(% per DI); Intercepts in the native units of each variable. $ = PD differs from CG (p < .05; two-sample t-test on slopes/intercepts).

Overall, these estimates indicate that PD participants start from a less favourable baseline (lower speed and step length, higher double support and step-length CV) while showing comparable DI-related sensitivity to controls for most outcomes; the only slope difference emerged for step length, which declined more steeply with increasing cognitive load in PD.

Table 5 reports the group-level parameters from linear fits between the DI and DTC for each gait outcome. For speed cost and step-length cost, people with PD showed both steeper positive slopes (speed: 9.75 ± 6.39%/DI; step length: 5.44 ± 3.67%/DI) and higher intercepts (speed: 4.88 ± 5.72%; step length: 1.86 ± 3.53%) than controls (all $, p<.05), indicating greater baseline cost and a faster rise in cost as cognitive load increased. By contrast, double-support cost and cadence cost exhibited similar slopes and intercepts in the two groups, with overlapping variability. For step-length CV cost, between-group differences were not reliable and dispersion was large, reflecting substantial inter-individual variability. Across outcomes, cost-DI correlations were positive and comparably strong in PD and controls (e.g., speed cost: r = 0.84 ± 0.19 in PD vs 0.79 ± 0.29 in CG), consistent with a monotonic increase in DTC as task difficulty rises and suggesting similar load sensitivity despite higher overall costs in PD for the key spatial measures.

	Slope		**Intercept**		**Correlation**	
	CG	PD	CG	PD	CG	PD
Speed Cost [%]	6.8 ± 5.09	9.75 ± 6.39$	1.77 ± 5.56	4.88 ± 5.72 $	0.79 ± 0.29	0.84 ± 0.19
Step length Cost [%]	3.41 ± 2.78	5.44 ± 3.67$	0.01 ± 2.08	1.86 ± 3.53 $	0.77 ± 0.36	0.85 ± 0.16
Double Sup. % GC Cost [%]	4.52 ± 4.59	5.66 ± 5.19	0.3 ± 3.59	1.85 ± 5.02	0.68 ± 0.5	0.69 ± 0.38
Cadence Cost [%]	3.74 ± 3.71	5.13 ± 4.08	1.73 ± 4.7	3.02 ± 4.1	0.7 ± 0.32	0.76 ± 0.31
CV Step length Cost [%]	15.96 ± 32.88	24.03 ± 36.11	1.13 ± 18.71	7.3 ± 31.38	0.28 ± 0.58	0.4 ± 0.56

Table 5. Group-level mean (± SD) slope, intercept, and correlation (r) from linear regressions of dual-task cost (DTC, %) on Difficulty Index (DI) in controls (CG) and Parkinson's disease (PD). Slope units: % cost per DI unit. Intercepts: % cost at DI = 0. Variables: Speed cost, Step-length cost, Double-support (%GC) cost, Cadence cost, step-length CV cost. $ = PD differs from CG (p < .05; between-group test on the parameter).

4.4 Discussion

This study examined a graded back-counting DT paradigm (-1, -3, and -7) in individuals with PD and age-matched controls, combining condition-specific analyses with a continuous DI. Three main observations emerged. First, under ST conditions, the PD group demonstrated the expected poorer baseline gait, slower speed, shorter steps, and greater double support compared to the control group (*Table 2*). Second, as cognitive demand increased, gait deteriorated in a dose-response manner: speed, step length, and cadence declined, while double support increased, with the most notable differences at DT-7 (*Table 2*; *Figure 1*). Third, continuous modelling of cognitive load revealed linear DI-gait relationships in both groups (*Figure 2*; *Table 4*,); the slopes were broadly similar between PD and controls, although intercepts (baselines) were consistently worse in PD. The only difference in slope appeared for step length, which declined more steeply with increasing DI in PD. The same pattern persisted when outcomes were expressed as DTC: PD showed higher baseline costs and steeper cost-DI slopes for speed and step length, with comparable cost-DI correlations across groups (*Table 5*,). Crucially, the magnitude of variance explained by the fixed factors for step length (R_m^2 = 0.206) and the very large effect size for its reduction under cognitive load (Cohen's d = 1.08 from ST to DT-7 in PD) highlight that this is not merely a statistical phenomenon, but a clinically meaningful deterioration. Disproportionate step shortening, such as the observed baseline deficit of nearly 7 cm in PD, which further degrades under DT, is a known biomechanical precursor to festinating gait and freezing episodes. Functionally, this indicates that when individuals with PD are distracted in everyday environments, they lose their adequate base of support much more rapidly than healthy peers. This translates directly into compromised dynamic balance and a significantly elevated risk of forward falls, demonstrating a functional penalty that goes well beyond the purely statistical group interaction.

These findings align with previous research showing that DT walking accentuates gait impairments in PD compared to ST and that increasing attentional demand utilises shared executive resources, diminishing gait automaticity (Al-Yahya et al., 2011; Caetano et al., 2019; Kelly et al., 2012; Lord et al., 2010; Raffegeau et al., 2019; Rochester et al., 2014; Yogev-Seligmann et al., 2008). By grading the subtraction task and measuring effective cognitive load with DI (net subtraction rate including errors), the present study reduces heterogeneity caused by inter-individual numeracy and error propagation, which are common sources of variability for serial subtraction (Raffegeau et al., 2019; Srygley et al., 2009).

Recent proposals to rank or calibrate DT cognitive tasks support this approach (Almutairi et al., 2025). Our data expand the field by differentiating baseline differences (intercepts) from load sensitivity (slopes), demonstrating that once load is measured, changes in speed, step length, and cadence with rising difficulty are similar in PD and controls, while double support proportionally increases with DI as a stability-oriented adaptation. The lack of a consistent DI effect on step-length CV suggests that variability may be less directly influenced by moment-to-moment cognitive load during short overground trials, or that longer sampling periods or different cognitive stressors are required to produce clear changes. The cognitive and perceptual correlations support the interpretation of a graded, mainly cognitive burden.

While the observed declines in gait performance under DT conditions are widely interpreted as reflecting a loss of gait automaticity, it is necessary to explicitly state what can and cannot be inferred from the measured spatiotemporal variables. Spatiotemporal metrics such as speed, step length, and double support are macroscopic behavioural outputs. Therefore, they can be used to infer the functional consequences of reduced automaticity: specifically, the degree to which walking relies on shared executive-attentional resources, as evidenced by the magnitude of performance deterioration under cognitive load. They also reflect the general compensatory strategies adopted by the central nervous system, such as increasing double support time to prioritise stability.

However, these variables cannot directly infer the underlying neurophysiological mechanisms of automaticity, such as the specific compensatory shift from basal ganglia circuits to cortical control networks. Furthermore, spatiotemporal data cannot reveal the precise biomechanical or motor-control adjustments, such as joint kinematics, muscle co-activation patterns, or changes in dynamic postural reflexes, that occur during motor-cognitive interference. Consequently, within the context of this study, "automaticity" is operationalised functionally as the resilience of the locomotor pattern against concurrent cognitive load, rather than as a direct neural or kinematic measurement.

Correct responses declined, and errors increased, with task difficulty, while DI increased stepwise from DT-1 to DT-7 without group differences (*Table 3*), indicating comparable effective loads for both PD and controls at each level. Clinically, the proportion of participants walking at less than 1.0 $m \cdot s^{-1}$ grew with difficulty, especially in PD, highlighting the risk of approaching a common safety threshold as cognitive demands increase. Perceived mental fatigue (VAS M) rose significantly with difficulty and more sharply in PD. In contrast, physical fatigue (VAS P) changed only modestly (*Table 3*, above), consistent with the idea that this paradigm primarily taxes executive attentional resources rather than physical effort. Functionally, this disproportionate rise in mental exhaustion helps explain the reduced daily physical activity often observed in PD. It suggests that patients may subconsciously avoid complex, multitasking environments in their daily lives, not because their muscles are physically tired, but to prevent the rapid cognitive burnout required to safely control their gait. Furthermore, dropping below the 1.0 $m \cdot s^{-1}$ safety threshold (as seen in 29 PD participants at DT-7) means that an individual may physically struggle to safely cross a street during a standard pedestrian light cycle or navigate crowded spaces independently, leading to progressive social isolation and loss of functional independence. These results align with and complement recent evidence that DT-focused and motor-cognitive interventions can improve DT gait performance in PD, although superiority over single-task training depends on task content, dosage, and transfer (Lin et al., 2024; Sarasso et al., 2024; Wong et al., 2023). By offering graded levels and a continuous load metric, the current protocol provides practical progression rules (e.g., −1→−3→−7) and sensitive readouts (DI-linked slopes/intercepts; DTC-DI relations) that are well suited for longitudinal monitoring and trials. Furthermore, the connection between executive functions and DT gait, as reported elsewhere, supports combining brief executive screening with graded DT testing to explain individual differences in DI-gait coupling (Caetano et al., 2019; Rochester et al., 2014; Yogev-Seligmann et al., 2008).

The strengths of the study include a graded DT design with randomised levels, a straightforward and interpretable DI that accurately reflects effective load, and complementary analyses on raw data and cost variables using linear mixed models. Additionally, the ecological testing in the on-medication state enhances real-world relevance. Limitations involve the cross-sectional nature of the study, dependence on a single cognitive task family (serial subtraction), utilisation of a 10-m walkway, which may underestimate variability metrics, and differences between groups in some questionnaire measures (such as education, mood, sleep, SF-36), which, although noted, could still influence results due to residual confounding. Future research should explore other executive-demanding tasks (e.g., set-shifting, inhibition), establish test-retest reliability and the minimal detectable change for DI-based outcomes, compare on versus off medication states, and assess the predictive validity for falls and functional decline. Work on additional DT-sensitive features (such as turning and gait smoothness) indicates that calibrated DT paradigms can also detect more subtle motor control deficits in PD (Caronni et al., 2025).

4.5 Conclusions

A graded, subtraction-based DT, combined with a DI, produces linear dose-response changes in gait in both PD and controls. PD individuals show worse baselines (lower speed and step length; higher double support and step-length CV) but generally similar load sensitivity to increasing cognitive demand; only step length exhibits a sharper decline with rising DI. The DI correlates with cognitive performance and perceived mental effort, and the proportion of walkers below 1.0 $m \cdot s^{-1}$ increases with clinically meaningful and actionable difficulty information for monitoring and progression. Overall, graded and calibrated DT paradigms provide a practical framework for examining motor-cognitive interference, enabling cross-sectional comparisons, and delivering sensitive long-term outcomes for rehabilitation and clinical research in PD.

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CHAPTER 5: Gait Training with an Intelligent Treadmill with Low and High-Interference Biofeedback in Parkinson's Disease: A Comparison of Effects on Dual-Task Gait Performance

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Abstract

Parkinson's disease (PD) is characterised by shorter stride length, slower walking speed, and increased double support, which become more pronounced under dual-task (DT) conditions. Treadmill training with real-time biofeedback (BF) may improve gait regulation, especially when the feedback is cognitively demanding and combined with higher metabolic load. This study assessed the safety and effectiveness of five 4-week aerobic protocols in PD, comparing traditional treadmill walking with low- and high-interference BF, with or without increased metabolic intensity.

Sixty-one community-dwelling individuals with idiopathic PD (67.3 ± 7.7 years; H & Y 1.8 ± 0.6) were randomly assigned to Control (C, n=12), traditional treadmill (T_T, n=12), low-interference biofeedback (BF_T, n=12), high-interference biofeedback (BF_HI_T, n=15), or high-interference plus higher metabolic intensity (BF_HI_HM_T, n=10). Treadmill groups trained 3× per week for 4 weeks (12 sessions, ~45 minutes each), progressing from 80% to 110-120% of self-selected speed; BF_HI_HM_T included incline and weighted vest to target ~70-100% respiratory compensation point. Gait (10-m walk) was evaluated in single-task (ST) and DT-1, DT-3, and DT-7 conditions; spatiotemporal parameters, dual-task costs (DTC), and a Difficulty Index (DI) were calculated. Aerobic fitness was assessed via maximal aerobic exercise testing. Linear mixed-effects models (Time, Group, Time×Group; REML, Satterthwaite degrees of freedom, Holm correction) and ANOVAs on change scores were used.

No Time or Time×Group effects emerged for $\dot{V}O_2max$, gas exchange threshold, or respiratory compensation point (all $p>0.05$); only peak power output differed between groups ($p = 0.041$). Gait speed showed no significant Time or Time×Group effects in any condition (all $p≥0.060$). Step length improved over time in ST, DT-3, and DT-7 (Time $p≤0.037$); at DT-7, a Time×Group interaction ($p = 0.009$) reflected a marked increase in BF_HI_HM_T (56.9 ± 4.08 to 64.09 ± 6.91 cm, $p = 0.003$), with a greater Δ than C (0.47 ± 4.27 cm, $p = 0.036$) and BF_T (−0.84 ± 5.04 cm, $p = 0.007$). Double support decreased across all conditions (Time $p≤0.025$); in DT-7, BF_HI_HM_T reduced from 34.11 ± 5.53 to 30.29 ± 5.82% ($p = 0.022$), with Δ differing from BF_HI_T (−2.65 ± 2.89 vs 0.95 ± 3.01%, $p = 0.019$). DT-7 speed and step length costs showed Time×Group interactions ($p = 0.042$ and 0.035), with the largest descriptive reductions in high-interference groups. DI decreased more in BF_HI_T than C at DT-1 (Δ −0.15 ± 0.14 vs 0.01 ± 0.09; $p = 0.021$) and DT-7 (Δ −1.92 ± 1.63 vs 1.25 ± 2.14; $p = 0.021$). Adherence exceeded 11/12 sessions in all groups ($p = 0.497$), with higher perceived exertion in BF_HI_HM_T (group $p = 0.033$). In conclusion, high-interference BF treadmill training, especially when combined with higher metabolic intensity, is feasible and selectively improves step length, double support, and DT difficulty in PD, despite no change in aerobic fitness over four weeks.

5.1 Introduction

Parkinson's disease (PD) is a progressive neurodegenerative disorder characterised by motor deficits such as bradykinesia, rigidity, postural instability, and gait impairment. Gait in PD often shows reduced stride length, slower speed, and increased double-support time, all of which contribute to diminished functional mobility and a higher risk of falls (Ambrus et al., 2019). Aerobic exercise is widely recognised as an effective non-pharmacological intervention in PD rehabilitation, enhancing cardiovascular fitness, motor performance, and neuroplasticity (Ahmad et al., 2024; Ferrazzoli et al., 2020).

Among aerobic modalities, treadmill walking has been widely studied for its potential to improve gait automaticity and motor adaptation. Studies indicate that treadmill walking can enhance the stride length-cadence relationship compared with overground walking, suggesting better internal gait regulation (Ambrus et al., 2019). Systematic reviews and meta-analyses further confirm that treadmill training, whether performed with incline, body-weight support, or feedback, produces beneficial effects on spatiotemporal gait parameters, such as stride length, cadence, and walking speed, in PD populations.(Bishnoi et al., 2022).

The integration of real-time biofeedback (BF) into treadmill training marks a promising advancement in PD gait rehabilitation. By providing instant visual or vibrotactile information about gait parameters (e.g., step length, cadence, propulsive force), BF aims to improve motor awareness and support motor learning (McMaster et al., 2022). For instance, visual feedback of anterior ground reaction force has been shown to enhance propulsive force generation and stride parameters in individuals with PD (Baudendistel et al., 2024). Similarly, wearable vibrotactile feedback systems have proven to be feasible and safe, resulting in measurable improvements in gait speed and step length (McMaster et al., 2022).

Despite these promising results, further research is needed to clarify how different types of BF affect gait and cardiorespiratory outcomes. In particular, the potential advantages of more cognitively demanding training protocols, such as those involving high-interference feedback, where step length and cadence cues are presented in randomised sequences, remain underexplored. Furthermore, combining high cognitive load with increased metabolic intensity, achieved through treadmill incline and external loading, could improve the training's overall impact by simultaneously engaging attention and promoting aerobic adaptation.

A key element of this study is the standardisation and individualisation of the training protocols. All treadmill-based interventions used self-selected walking speed as a reference point for initial intensity, with a programmed increase over the four-week period (from 80% to 120% of self-selected speed). In the group with higher metabolic intensity (incline and weighted vest), the same speed progression was applied, while treadmill grade and external load were gradually increased to elicit a physiological workload approaching each participant's respiratory compensation point (RCP), estimated through standard equations for treadmill walking VO_2. Furthermore, BF targets (step length and cadence) were personalised using normative reference data adjusted for sex, age, and body height, according to the interactive model proposed by Moe-Nilssen and Helbostad (Moe-Nilssen and Helbostad, 2020) .

The aim of this study is to assess the effectiveness of treadmill-based gait training with real-time BF in individuals with PD, in comparison to traditional treadmill walking without feedback. Specifically, we examine whether the integration of high-interference BF (randomly alternating step length and cadence cues within each session) and the addition of high metabolic intensity (through treadmill incline and weighted vest) result in greater improvements in gait parameters, both under single-task and DT conditions, as well as in aerobic fitness. We hypothesise that BF-based training will produce better outcomes than traditional treadmill training and that combining attentional demand with increased intensity will enhance these benefits.

5.2 Materials and Methods

5.2.1 Participants

Sixty-one community-dwelling older adults with idiopathic PD (19 females; 67.3 ± 7.7 years; 1.8 ± 0.6 H & Y) were enrolled in the study. Participants were recruited from the local community through flyers, advertisements, word of mouth, and clinical referrals after preliminary screening.

Inclusion criteria were: diagnosis of PD according to the UK Parkinson's Disease Society Brain Bank clinical diagnostic (Hughes et al., 1992); Hoehn and Yahr stage < 3.0 (Hoehn and Yahr, 1967), no orthopaedic injuries in the past six months; no acute neurological events; no lower-limb pain that limits walking; no severe psychiatric or medical conditions; no cognitive impairment, defined as Mini-Mental State Examination (MMSE) > 24 or Montreal Cognitive Assessment (MoCA) > 16 (Hoops et al., 2009; Ramaker et al., 2002).

Exclusion criteria included: presence of deep-brain stimulation or a pacemaker, major depression, unstable health conditions, or sensory deficits that interfere with assessments. All participants had normal or corrected-to-normal vision and could follow instructions during testing and training.

Participation in the experiment was voluntary with informed consent in accordance with the Ethics Committee for Approval of Research on Humans (CARU) of the University of Verona (n. 08/2022).

PD diagnosis and disease staging were confirmed by a trained neurologist. The average disease duration was 5.2 ± 3.7 years. Testing was conducted during the "on" phase of dopaminergic medication; however, the time from last intake was not standardised due to individual differences in therapy, in line with previous research (Salazar et al., 2017).

Functional mobility was assessed using the Short Physical Performance Battery (SPPB) (Guralnik et al., 1994), and quality of life was assessed using the SF-36 questionnaire, which provides norm-based Physical (PCS) and Mental (MCS) Component Summary scores (Apolone and Mosconi, 1998; Garratt and Stavem, 2017). Additional psychological assessments included the State-Trait Anxiety Inventory (STAI-Y I and II) (Bieling et al., 1998; Linde et al., 2022; Wang and Gorenstein, 2013), the Beck Depression Inventory-II (BDI-II) (Gellman and Turner, 2013), and the Pittsburgh Sleep Quality Index (PSQI) (Buysse et al., 1989).

After screening, participants were randomly assigned to five experimental groups:

- Control group (C): 12 participants (5 females; 69.8 ± 7.8 yrs; 1.8 ± 0.6 H & Y)
- Traditional treadmill training (T_T): 12 participants (5 females; 65.9 ± 6.5 yrs; 1.9 ± 0.8 H & Y)
- Low-interference biofeedback treadmill training (BF_T): 12 participants (3 females; 66.5 ± 8.1 yrs; 2.0 ± 0.8 H & Y)
- High-interference biofeedback treadmill training (BF_HI_T): 15 participants (3 females; 66.3 ± 11.3 yrs; 2.0 ± 0.6 H & Y)
- High-interference biofeedback treadmill training with higher metabolic intensity (BF_HI_HM_T): 10 participants (3 females; 63.2 ± 7.4 yrs; 2.0 ± 0.8 H & Y), reduced from an initial 12 due to two dropouts (one for scheduling conflicts, one for difficulty completing training).

5.2.2 Exercise intervention procedure

5.2.2.1 Training protocols

All treadmill-based training groups followed the same number of training sessions, session durations, and a standardised progression of walking speed tailored to each participant's abilities. In all groups receiving biofeedback (BF_T, BF_HI_T, BF_HI_HM_T), feedback was provided on both step length and cadence. The specific modalities of BF delivery varied by training type and are detailed in the following sections.

For the groups with BF, the progression of treadmill speed and feedback targets followed two predefined schemes based on disease severity and participants' ability to walk on the treadmill. For participants classified as H & Y stage 1-1.5, the progression involved walking at 80% of self-selected speed during the first four sessions, 100% during sessions 5-8, and 120% of self-selected speed during sessions 9-12. For those in stage 2-2.5, the progression was 80% during the first four sessions, 90% during sessions 5-8, and 110% during sessions 9-12.

Target values for step length and cadence were customised based on normative data adjusted for sex, age, and height (Moe-Nilssen and Helbostad, 2020), with gradual increases over the 4-week training period. Perceived exertion was monitored using the Borg scale CR-10 (Borg, 1998).

Training sessions were carried out using three different treadmill systems, depending on laboratory and equipment availability:

- h/p/cosmos mercury med treadmill with integrated OptoGait system (h/p/cosmos sports & medical gmbh, Nussdorf-Traunstein, Germany);
- Run Race treadmill with externally mounted OptoGait system (version 1.13.24.0) (Technogym, Cesena, Italy; Microgate, Bolzano, Italy);
- SPLIT-BELT treadmill with externally mounted OptoGait system (version 1.13.24.0) (WOODWAY GmbH, Weil am Rhein, Germany; Microgate, Bolzano, Italy).

A 48-hour recovery period was necessary between each session. To minimise any time-related physiological changes in participants, all sessions were completed within 6 weeks.

All participants completed a supervised aerobic training programme consisting of three sessions per week, each lasting about 45 minutes, totalling 12 sessions over four weeks. Each session consisted of three 10-minute exercise bouts, separated by 5 minutes of passive rest. All training was carried out in a laboratory setting under the supervision of an exercise specialist.

Control participants were instructed to maintain their normal lifestyle throughout the study period, with no specific guidelines for physical activity. Let's now see each protocol in detail:

Traditional Treadmill Training (T_T):

Participants walked on a motorised treadmill without any form of feedback. The structure, number of sessions, and progression of walking speed followed the same personalised protocol applied across all treadmill-based training groups, with adjustments based on disease stage and treadmill walking ability. No visual or cognitive cues were provided.

Low-Interference Biofeedback Treadmill Training (BF_T):

This condition mirrored the general structure but included real-time visual feedback on either step length or cadence, delivered via the OptoGait system (version 1.13.24.0, Microgate, Bolzano, Italy).

Feedback was shown using a colour-coded "traffic-light" interface (green = correct, orange = slightly off-target, red = incorrect) on a monitor placed in front of the participant (see *Figure 1*). Each training session concentrated on a single gait parameter, alternating between sessions to maintain a 50% distribution of cadence and step length feedback throughout the entire intervention.

High-Interference Biofeedback Treadmill Training (BF_HI_T):

Participants in this group followed the same overall protocol as the BF_T group. However, within each session, the visual feedback alternated randomly between step length and cadence, maintaining an approximately 50/50 split across total training time. This random alternation introduced greater contextual interference.

High-Interference Biofeedback Treadmill Training with Higher Metabolic Intensity (BF_HI_HM_T):

This protocol combined the randomised feedback alternation of the BF_HI_T group with increased metabolic workload. Treadmill incline and external load (weighted vest) were progressively adjusted to attain training intensities of approximately 70%, 80%, and 100% of each participant's Respiratory Compensation Point (RCP). The total effective body mass (body weight plus added load) was used to estimate oxygen consumption using the standard ACSM treadmill walking equation.

$$VO_2 = (0.1 \times v) + (1.8 \times v \times G) + 3.5$$

where v is walking speed ($m \cdot min^{-1}$) and G is fractional treadmill grade (e.g., 5% = 0.05). Adjusted VO_2 values were normalised to total effective mass to determine the corresponding percentage of RCP for all participants. The same progression schedule (80%, 95%, 110% of predicted gait parameters) was applied to feedback targets as in the BF_T and BF_HI_T groups.

All sessions were supervised to ensure adherence and safety. Participants were continuously monitored for signs of fatigue, instability, or discomfort. During each training session, metabolic intensity was assessed at every change in treadmill speed using a metabolic cart, in order to verify that the achieved metabolic load matched the prescribed target intensity.

The BF groups watched their gait parameter (i.e., step length or cadence) on the screen and attempted to perform the target requested for that cognitive feedback as accurately as possible. The software calculated the orange and red boundaries at 5% and 15% of the correct target (see *Figure 1, Figure 2*).

Figure 1. The interface of the OptoGait software of the step length Biofeedback target.

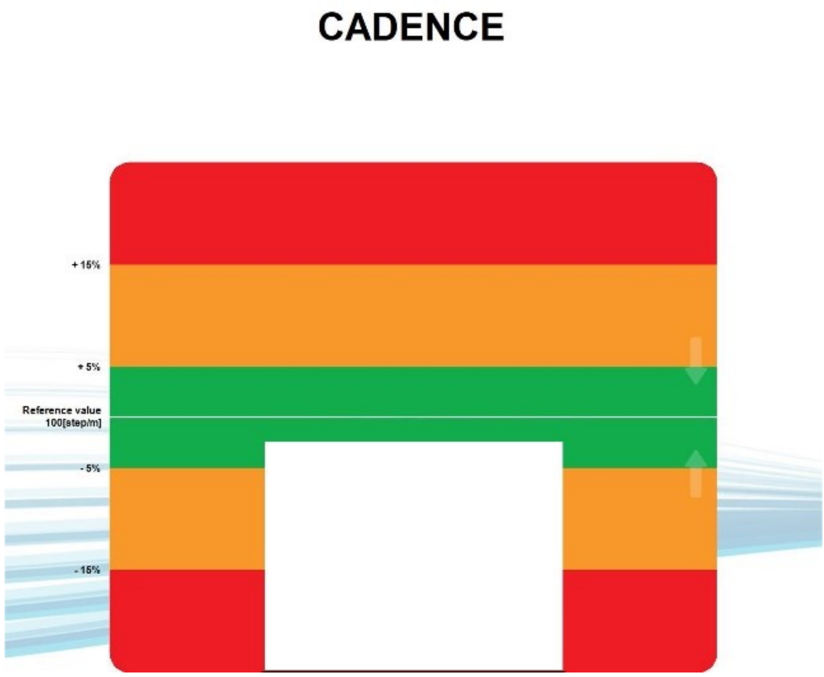

Figure 2. The interface of the OptoGait software of the cadence Biofeedback target.

5.2.3 Tests and Calculations

Participants attended eighteen study sessions, conducted by trained investigators at a consistent time of day (10:00 a.m. ± 2 hours) in environmentally controlled laboratories. Sessions took place either in the exercise physiology laboratory at the University of Verona (temperature: 22-25 °C; relative humidity: 55-65%) or in the Microgate motion-analysis laboratory (Bolzano, Italy). The same operators carried out all assessments and training at both locations, maintaining identical environmental conditions, equipment settings, and procedural standards. A 48-hour recovery period was required between sessions. To minimise time-related physiological variability, all sessions were completed within six weeks.

Baseline assessments (PRE) were carried out over three separate days. On the first day, participants completed structured interviews and underwent anthropometric and clinical evaluations. The recorded variables included limb dominance and side of symptom onset, education (years), and Hoehn and Yahr stage (H&Y) (Hoehn and Yahr, 1967). Duration of PD (in years since diagnosis) and severity of motor symptoms as measured by the Unified Parkinson's Disease Rating Scale - Part III (UPDRS-III) (Goetz et al., 2004) .

Cognitive functioning was assessed using the Mini-Mental State Examination (MMSE) and the Montreal Cognitive Assessment (MoCA) (Hoops et al., 2009). Body height and weight were measured using a vertical stadiometer and digital scale (Seca, Leicester, UK), and these measurements were used to calculate the BMI. Functional mobility was evaluated using the Short Physical Performance Battery (SPPB) (Guralnik et al., 1994), and fall risk was estimated with the Timed Up and Go Test (TUGT) (Huang et al., 2011) . Participants also completed the Short Form-36 Health Survey (SF-36) (Apolone and Mosconi, 1998), from which the Physical (PCS) and Mental (MCS) Component Summary scores were derived (Garratt and Stavem, 2017).

Psychological and behavioural status were measured using the State-Trait Anxiety Inventory Form Y-I and Y-II (STAI-Y) (Linde et al., 2022; Spielberger et al., 1983), the Beck Depression Inventory-II (BDI-II) (Wang and Gorenstein, 2013), and the Pittsburgh Sleep Quality Index (PSQI) (Curcio et al., 2013).

The second session focused on gait assessment using the 10-Metre Walk Test (10MWT) in both single-task and DT conditions. DT trials were administered with three levels of cognitive difficulty, serial subtraction by −1, −3, and −7 (i.e., DT-1, DT-3, DT-7), consistent with previous work pairing mobility tests with serial subtraction tasks (Al-Yahya et al., 2011; Brustio et al., 2017).

On the third day, participants underwent a Maximal Aerobic Exercise Test on a cycle ergometer to assess their aerobic capacity ($\dot{V}O_{2max}$ [$ml\cdot min^{-1}\cdot kg^{-1}$]); and respiratory thresholds (GET [$ml\cdot min^{-1}\cdot kg^{-1}$]; RCP [$ml\cdot min^{-1}\cdot kg^{-1}$]).

5.2.3.1 Dual-Task Walking Assessment

During the second study session, participants performed four walking conditions in randomised order:

- Single-task (ST): walking at a self-selected pace without concurrent tasks.
- Dual-task 1 (DT-1): walking while counting backwards by ones.
- Dual-task 3 (DT-3): walking while counting backwards by threes.
- Dual-task 7 (DT-7): walking while counting backwards by sevens.

Each condition was performed twice, resulting in eight trials per participant. Walking trials took place on a 10-metre walkway, with participants asked to start walking 1.5 metres before the sensor area and continue for 1.5 metres beyond its end to ensure steady-state gait was captured. Participants were instructed to keep a self-selected pace throughout.

Spatiotemporal gait parameters were measured using a validated photoelectric system (OptoGait, version 1.13.24.0, Microgate, Bolzano, Italy) integrated with ten one-metre photoelectric bars and dedicated software. The system automatically recorded the time taken to traverse the central 10-metre segment, speed ($m \cdot s^{-1}$), cadence (steps/min), step length (cm), and double support (percentage of the gait cycle). The faster trial for each condition was retained for analysis.

In DT conditions, participants were instructed to count aloud while walking, using subtraction tasks initiated from a random two-digit number between 51 and 99. They were explicitly told not to prioritise either task (walking or counting). The experimenter recorded the number of correct responses and errors. After each trial, participants rested in a seated position for approximately 2 minutes or until they verbally confirmed full recovery, after which they proceeded to the next condition.

5.2.3.2 Maximal aerobic exercise test

The Maximal Aerobic Exercise Test was used to assess participants' cardiorespiratory fitness levels and was performed on an electronically-braked cycle ergometer (Sport Excalibur, Lode, Groningen, NL). Respiratory variables were continuously recorded breath-by-breath throughout the test using a metabolic cart (Quark B2, Cosmed, Roma, Italy), which was calibrated before each testing session according to the manufacturer's instructions. Heart rate (HR) was monitored continuously. Before each visit, participants were instructed to arrive at least two hours after fasting, avoid alcohol and vigorous exercise for at least 24 hours beforehand, and abstain from caffeine on the morning of the visit. The height and depth of the handlebars and seat of the cycle ergometer were individually adjusted for each participant before the test and kept the same for reassessment after the 4-week training period. All participants completed a medically supervised step-ramp test protocol as follows:

a 6-minute constant work rate bout at a workload stage of 55 watts for males (40 W for females); a 4-minute rest period without pedalling; and a ramp-incremental bout where the workload increased by 15 W each minute from 0 W for males (10 W/min for females) until maximal effort criteria were met: Respiratory quotient > 1.1 and heart rate exceeding 90% of the predicted maximum based on age (American College of Sports Medicine, 2000), or until participants could no longer maintain the required workload (see *Figure 3*). The protocol required a five-minute resting period before measuring HR, blood pressure, and aerobic capacity. Blood pressure was continuously measured every two minutes during the test. The highest completed workload was termed Power Output Peak (POPeak), and maximal oxygen consumption ($\dot{V}O2max$) was calculated as the average of the values recorded over the last 10 seconds of exercise. Furthermore, the individual first and second ventilatory thresholds, called gas-exchange threshold (GET) and respiratory compensation point (RCP), were detected by visual inspection based on ventilatory equivalents and end-tidal fractions of O2 and CO2. (Wasserman, 1986), by three independent investigators. The PO corresponding to the RCP was used to prescribe the exercise for the CT group. Finally, all ventilatory variables were expressed for the weight of each participant, recording the following variables: $_{Rel}\dot{V}O_{2max}$ [$ml \cdot min^{-1} \cdot kg^{-1}$]; $_{Rel}GET$ [$ml \cdot min^{-1} \cdot kg^{-1}$]; $_{Rel}RCP$ [$ml \cdot min^{-1} \cdot kg^{-1}$].

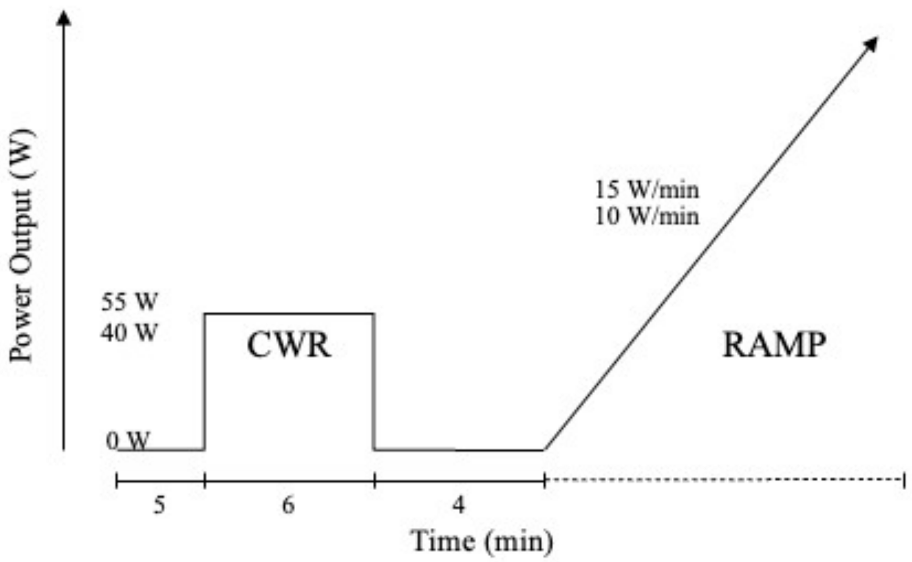

Figure 1. The procedure of the maximal aerobic exercise test: the step-ramp test.
Note: CWR indicates constant work rate step; Ramp indicates the incremental ramp test.

5.2.4 Calculations

The effect of DT on gait performance was expressed in two ways.

The first one was via DTC: for each spatiotemporal parameter, the relative cost of performing the DT compared with the single-task condition was calculated using the formula:

$$DTC = \frac{-(DT - ST)}{ST} \times 100$$

A minus sign was placed before the numerator in the subtraction to facilitate data interpretation. This convention ensures that positive values reflect greater costs during DT compared to single-task walking, indicating gait deterioration induced by the cognitive task (Liu et al., 2018).

The second way was via DI: We created an additional variable to quantify the cognitive difficulty. We named this variable the Difficulty Index (DI), and we computed it as the reciprocal of the number of correct responses per second for each DT condition, using the following formula:

$$DI = \frac{1}{\frac{Number\ of\ response - number\ of\ errors}{Time\ (s) over\ 10m}}$$

Under cognitive load, the number of correct responses in one second will likely decrease and, as a result, the DI will increase, reflecting the relative cognitive challenge of the subtraction tasks coupled with walking.

5.2.5 Statistical analysis

The sample size was determined through an a priori power analysis to identify the number of participants required per group, assuming a medium effect size (ES). Power analysis was performed using G*Power 3.1 (test family: F tests; statistical test: repeated measures ANOVA, with a within-between interaction). We aimed for a medium effect (Cohen's $f = 0.25$), $\alpha = .05$, power $(1-\beta) = .80$, 5 groups, 2 measurements (PRE, POST), assuming a correlation among repeated measures $r = .50$ and $\varepsilon = 1.00$. The analysis showed a total sample size of $N = 55$ (about 11 per group) to detect the Time × Group interaction (critical $F = 2.557$).

The final sample ($N = 61$) exceeded the a priori requirement ($N = 55$), thereby meeting the planned ≥80% power to detect a medium-sized Time × Group interaction under the specified assumptions (G*Power output: actual power ≈ .821 at $N = 55$). Statistical analysis was conducted using Jamovi software (version 2.6.44; https//www.jamovi.org, Sydney, AUS). Numerical values were expressed as a mean with standard deviations (mean ± SD) for all parameters. Demographical and clinical variables were analysed using descriptive statistics. The one-way ANOVA was employed to assess differences between groups at PRE.

For all outcome variables, we employed linear mixed effects models (LMMs) to evaluate the effects of time (PRE vs. POST), group (C, T_T, BF_T, BF_HI_T, BF_HI_HM_T), and their interaction (Time × Group). This approach was specifically chosen to handle observed baseline imbalances between groups. By focusing on the Time × Group interaction, the LMM evaluates differences in the trajectory of change over time rather than comparing absolute values, effectively adjusting for initial disparities. Furthermore, each model incorporated participant ID as a random intercept to account for within-subject variability (Gueorguieva and Krystal, 2004).

LMMs were estimated using restricted maximum likelihood (REML) and Satterthwaite approximation for degrees of freedom. The models were fitted using the bobyqa optimiser. This method enabled the inclusion of participants with incomplete data and provided valid estimates under the assumptions of missing completely at random (MCAR) or missing at random (MAR).

Omnibus tests assessed the main effects of Time and Group, as well as their interaction. Where appropriate, post hoc pairwise comparisons were carried out using estimated marginal means, with Holm correction for multiple comparisons. Effect sizes were derived from model contrasts.

All analyses were conducted using an intention-to-treat approach and involved all available data from participants with baseline assessments (N = 61). A significance threshold of $p < 0.05$ was applied.

To compare Δ (POST - PRE) across groups, we conducted a one-way ANOVA. Normality of Δ values was checked with the Shapiro-Wilk test, and homogeneity of variances was assessed with Levene's test. When both assumptions were satisfied, we used the standard one-way ANOVA with Tukey post-hoc tests. If normality was met but variance homogeneity was violated, we applied Welch's ANOVA with Games-Howell post-hoc tests (for unequal variances). When normality was not satisfied, we employed the Kruskal-Wallis test for non-parametric data. Statistical significance was set at $p < 0.05$ (two-tailed); p-values from post-hoc tests were additionally adjusted using the Holm method to control the familywise error rate.

5.3 Results

Most baseline characteristics were similar across groups; however, significant differences between groups appeared for peak power output (PPO) and motor severity (measured by the UPDRS-III). The BF_HI_HM_T group exhibited significantly higher PPO compared to the Control ($p = 0.048$) and had lower UPDRS-III scores, indicating milder motor impairment compared to the Control ($p = 0.032$) and BF_T ($p = 0.015$) groups.

Table 1 summarises participant characteristics and baseline group comparisons.

(**Mean** ± *SD*)	**CO**	**T_T**	**BF_T**	**BF_HI_T**	**BF_HI_HM_T**	**Total**	p
N°	12	12	12	15	10	**61**	-
Age [yr]	**69.8** ± 7.8	**65.9** ± 6.5	**66.5** ± 8.1	**66.3** ± 11.3	**63.2** ± 7.4	**67.3** ± 7.7	0.481
Females # (%)	5 (42)	5 (42)	3 (25)	3 (20)	3 (33)	19 (31)	-
BMI [$kg \cdot m^{-2}$]	**26.2** ± 4.5	**25.5** ± 4.8	**25.4** ± 4.5	**27.9** ± 5.3	**24.7** ± 5.5	**26.0** ± 4.8	0.715
$Vo_{2\ max}$ [$mL \cdot min^{-1} \cdot kg^{-1}$]	**1778.9** ± 490.7	**1818.8** ± 506.4	**2019.2** ± 676.4	**2158.5** ± 676.4	**2549.6** ± 668.1	**2041.0** ± 640.9	0.121
PPO [w]	**130.0** ± 48.17	**136.1** ± 45.46	**152.3** ± 57.7	**169.4** ± 65.02	**206.1** ± 71.7 *	**156.3** ± 61.6	0.004
Education [yr]	**11.1** ± 3.3	**13.7** ± 5.1	**11.8** ± 4.3	**12.1** ± 3.41	**12.9** ± 2.9	**11.8** ± 4.3	0.385
H&Y	**1.8** ± 0.6	**1.9** ± 0.8	**2.0** ± 0.8	**2.0** ± 0.6	**1.9** ± 0.6	**1.8** ± 0.6	0.663
UPDRS-III	**34.1** ± 13.2	**34.0** ± 14.5	**35.8** ± 13.3	**25.1** ± 5	**20.3** ± 6 * +	**34.0** ± 12.6	0.001
PD duration [yr]	**5.3** ± 3.9	**4.4** ± 3.1	**5.1** ± 3.6	**5.0** ± 3.2	**5.2** ± 5.3	**5.2** ± 3.7	0.896
Hand dominant (right/left)	12 / 0	12 / 0	12 / 0	14 / 1	10 / 1	59 / 2	-
Side of impairment (right/left)	6 / 6	7 / 5	7 / 5	10 / 5	6 / 4	36 / 25	-
MMSE	**27.1** ± 1.3	**27.6** ± 1.8	**26.4** ± 2.3	**27.4** ± 1.7	**27.6** ± 2.2	**27.1** ± 1.8	0.632
MoCA	**26.6** ± 2.2	**26.1** ± 3.0	**26.4** ± 1.9	**23.7** ± 3.4	**25.4** ± 2.6	**25.6** ± 2.8	0.142
STAI - Y I	**35.9** ± 7.4	**32.3** ± 8.4	**33.9** ± 8.3	**32.6** ± 5.5	**35.5** ± 9.2	**34.1** ± 8.7	0.791
STAI - Y II	**38.1** ± 10.1	**38.3** ± 13.1	**33.9** ± 6.8	**32.3** ± 7.2	**41.1** ± 11.6	**36.4** ± 10.0	0.296
BDI-II	**8.8** ± 6.6	**9.2** ± 9.8	**4.9** ± 4.0	**6.4** ± 4.5	**11.5** ± 10.5	**8.3** ± 7.6	0.450
PSQI	**7.3** ± 3.7	**6.0** ± 2.8	**6.1** ± 1.1	**5.4** ± 3.7	**6.9** ± 3.7	**6.4** ± 3.3	0.627
TUGT [s]	**7.4** ± 1.4	**7.2** ± 1.4	**7.0** ± 1.7	**6.9** ± 1.3	**6.1** ± 0.7	**6.9** ± 1.3	0.329
SPPB	**11.1** ± 1.7	**11.3** ± 1.3	**11.9** ± 0.3	**11.8** ± 0.4	**11.9** ± 0.3	**11.6** ± 1.1	0.565
SF-36 PCS	**45.3** ± 8.0	**45.3** ± 8.3	**47.3** ± 7.1	**48.6** ± 5.5	**45.7** ± 8.5	**46.4** ± 7.4	0.813
SF-36 MCS	**42.7** ± 6.4	**43.5** ± 11.6	**47.7** ± 6.0	**50.0** ± 6.9	**42.6** ± 10.1	**45.3** ± 8.7	0.085

*Table 1. Mean ± SD of demographic, anthropometric, clinical, cognitive, and questionnaire variables are reported for each group (CO, T_T, BF_T, BF_HI_T, and BF_HI_HM_T) and for the total sample. Between-group comparisons were conducted using one-way ANOVA for parametric variables or the Kruskal-Wallis test for non-parametric data. Statistical symbols indicate significant differences between groups: * $p < 0.05$ vs. CO; # vs. T_T; + vs. BF_T; ^ vs. BF_HI_T.*

Table 2 presents data on training adherence, perceived exertion, and walking speed. In brief, training adherence remained consistently high across all groups, with no significant differences in the number of sessions completed (p = 0.497). On average, participants completed more than 11 out of 12 sessions in each group. No significant differences between groups were observed in mean treadmill walking speed during training (p = 0.604), with average values ranging from 1.3 to 1.4 $m \cdot s^{-1}$ across the treadmill-based conditions.

A significant group effect was found for perceived exertion measured by the Borg scale (p = 0.033). Participants in the BF_HI_HM_T group showed a significantly greater value of exertion levels compared to those in the T_T (p = 0.058, ES= 1.10) and BF_T (p = 0.051, ES= -1.15) groups.

(Mean ± *SD*)	**T_T**	**BF_T**	**BF_HI_T**	**BF_HI_HM_T**	p
N° training presence	11.6 ± 0.8	11.8 ± 0.4	11.5 ± 0.8	11.4 ± 0.9	0.497
Mean BORG training	3 ± 0.4	3 ± 1.4	3.4 ± 1.1	4.2 ± 1	0.033
Mean speed training [m·s-1]	1.3 ± 0.2	1.4 ± 0.2	1.4 ± 0.2	1.3 ± 0.1	0.604

Table 2. Adherence, perceived exertion, and walking speed during training sessions across groups: Mean ± SD values are reported for the number of training sessions attended (maximum = 12), mean perceived exertion during training (Borg scale), and average walking speed ($m \cdot s^{-1}$) for all treadmill-based groups. Between-group comparisons were performed using one-way ANOVA for normally distributed variables and the Kruskal-Wallis test for non-parametric data.

5.3.1 Aerobic Fitness

The linear mixed model revealed no significant main effect of time on any of the aerobic fitness outcomes, including maximal heart rate, peak power output (PPO), relative maximal oxygen uptake (VO_2max), gas exchange threshold (GET), and respiratory compensation point (RCP) (p > 0.05 for all comparisons). No significant interaction effects between time and group were observed for these variables (p > 0.05), indicating that the extent of change over the 4-week training period did not differ significantly among intervention groups. A main effect of group was only detected for PPO (p = 0.041), suggesting a difference in absolute power output between groups that was independent of time. However, no group effect was found for the other physiological parameters (p > 0.1) (*Table 3*).

(Mean ± SD) Δ POST - PRE	**Control**		**T_T**		**BF_T**		**BF_HI_T**		**BF_HI_HM_T**				
	PRE	**POST**	**PRE**	**POST**	**PRE**	**POST**	**PRE**	**POST**	**PRE**	**POST**	*Time*	*Gr.*	*x*
HR max [beats*min^{-1}]	131.25 ± 20.73	132.41 ± 22.47	138.03 ± 20.1	133.39 ± 24.66	130.48 ± 32.63	127.38 ± 31.67	139.67 ± 32.63	137.5 ± 28.12	141 ± 16.38	139.33 ± 16.5	0.888	0.836	0.606
	1.16 ± 9.01		-4.64 ± 13.58		-3.1 ± 6.47		-10.27 ± 33.28		-1.36 ± 3.56			0.725	
Power [W]	130 ± 48.17	130.49 ± 46.32	136.14 ± 45.46	137.58 ± 48.54	152.33 ± 57.7	151.02 ± 61.22	169.46 ± 65.02	169.33 ± 68.61	206.11 ± 71.7	203.67 ± 70.87	0.635	0.041	0.934
	0.49 ± 14		1.43 ± 8.77		-1.32 ± 11.03		-11.4 ± 39.66		-2 ± 9.03			0.948	
$Vo_{2\,max}$ [mL · min^{-1} · kg^{-1}]	25.07 ± 4.91	24.82 ± 5.44	27.09 ± 5.14	29.24 ± 5.46	25.82 ± 6.31	26.54 ± 6.64	28.07 ± 6.31	29.6 ± 9.69	33.87 ± 10.45	33.62 ± 11.38	0.098	0.130	0.495
	-0.25 ± 1.28		0.72 ± 2.14		0.63 ± 1.3		-1.15 ± 10.04		-0.21 ± 1.71			0.335	
GET [mL · min-1 · kg-1]	14.46 ± 2.6	15.12 ± 2.71	15.46 ± 3.11	14.94 ± 3	16.12 ± 3.5	16.03 ± 3.37	18.67 ± 3.5	19.26 ± 6.16	22.68 ± 7.17	22.82 ± 8.22	0.275	0.732	0.115
	0.67 ± 1.29		-0.51 ± 1		-0.09 ± 0.78		-0.71 ± 5.56		0.11 ± 1.61			0.114	
RCP [mL · min-1 · kg-1]	21.39 ± 3.45	21.67 ± 3.73	21.76 ± 4.84	21.93 ± 4.84	22.74 ± 4.8	22.77 ± 4.83	24.46 ± 4.8	25.58 ± 8.2	29.75 ± 9.45	30.18 ± 10.4	0.176	0.543	0.723
	0.27 ± 0.79		0.16 ± 1.15		0.03 ± 0.88		-0.88 ± 8.34		0.35 ± 1.62			0.443	

Table 3. Maximal aerobic test outcomes (mean ± SD) at baseline (PRE) and after the intervention (POST) for each group (Control, T_T, BF_T, BF_HL_T, BF_HLM_T). The right columns report p-values from the linear mixed model (LMM) for the main effects of Time, Group (Gr.), and Time × Group interaction (X). Between-group differences in POST–PRE changes (Δ) were analysed using one-way ANOV

5.3.2 Spatiotemporal Gait Parameters

Corresponding PRE, POST, and Δ values for each group and condition are displayed in *Figure* 4. No significant Time × Group interactions were observed for gait speed in any condition (p = ST: 0.576; DT-1: 0.164; DT-3: 0.318; DT-7: 0.086). The main effect of Time was not significant in ST (p = 0.060), DT-1 (p = 0.142), and DT-7 (p = 0.090), and showed a non-significant trend in DT-3 (p = 0.062). There were no Group effects (all p ≥ 0.350).

Regarding step length, a significant main effect of Time was observed in ST (p = 0.005), DT-3 (p = 0.037), and DT-7 (p < 0.001), whereas DT-1 did not reach significance (p = 0.071). A Group effect was present in ST (p = 0.033) but not in the other conditions (DT-1: p = 0.343; DT-3: p = 0.215; DT-7: p = 0.260). The Time × Group interaction was significant in DT-7 (p = 0.009) and not in ST, DT-1, or DT-3 (all p ≥ 0.368). In DT-7, BF_HI_HM_T showed a significant within-group increase (PRE 56.9 ± 4.08; to POST 64.09 ± 6.91, p = 0.003) and higher Δ values compared with CO (Δ BF_HI_HM_T 6.54 ± 6.29, Δ CO 0.47 ± 4.27; p = 0.036, ES = -1.285) and BF_T (Δ BF_HI_HM_T 6.54 ± 6.29, Δ BF_T -0.84 ± 5.04; p = 0.007, ES= 1.53). All other between-group contrasts for step length were not significant.

For double support, the main effect of Time was significant across all conditions (p = ST: 0.002; DT-1: 0.025; DT-3: 0.010; DT-7: 0.008). A Group effect was observed in ST (p = 0.009; Δ 0.010), but not in the DT conditions (DT-1: p = 0.408; DT-3: 0.623; DT-7: 0.374). The Time × Group interaction was significant in ST (p = 0.011) and DT-7 (p = 0.013), but not in DT-1 and DT-3 (both p > 0.10). In ST, PRE BF_HI_T showed a lower value than CO (PRE BF_HI_T 27.04 ± 3.03, PRE CO 32.20 ± 2.7, p = 0.034). Additionally, BF_HI_HM_T exhibited a lower POST value than CO (POST BF_HI_HM_T 25.58 ± 4.28, POST CO 31.66 ± 3.81; p = 0.016). In DT-7, BF_HI_HM_T demonstrated a significant within-group reduction (PRE 34.11 ± 5.53, POST 30.29 ± 5.82 p = 0.022) and a Δ that differed from BF_HI_T (p = 0.012).

Δ BF_HI_HM_T is significantly different to BF_HI_T (Δ BF_HI_HM_T -2.65 ± 2.89, Δ BF_HI_T 0.95 ± 3.01 p = 0.019 ES = 1.37). All remaining contrasts are detailed in *Figure 4*.

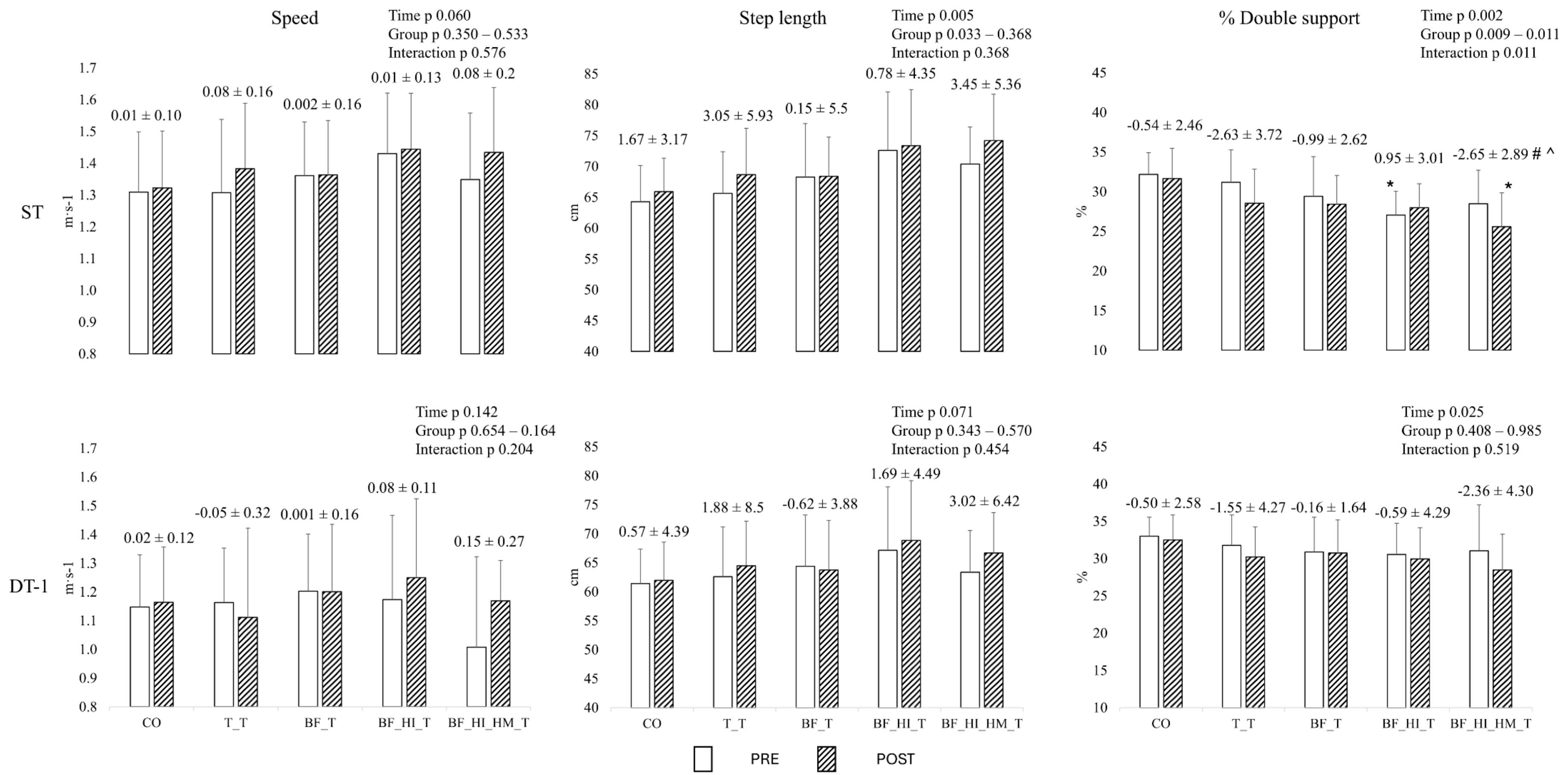

Figure 2; Panel A. For each group (C, T_T, BF_T, BF_HI_T, BF_HI_HM_T), PRE and POST values and within-group change (Δ) are reported under Speed ($m \cdot s^{-1}$),
Step length (cm), and Double support (DS, % gait cycle) in single task (ST) and first dual task conditions (DT-1).
*Symbols indicate significant comparisons: ~ = PRE-POST within-group difference; * vs CO; # vs T_T; + vs BF_T; ^ vs BF_HI_T.*

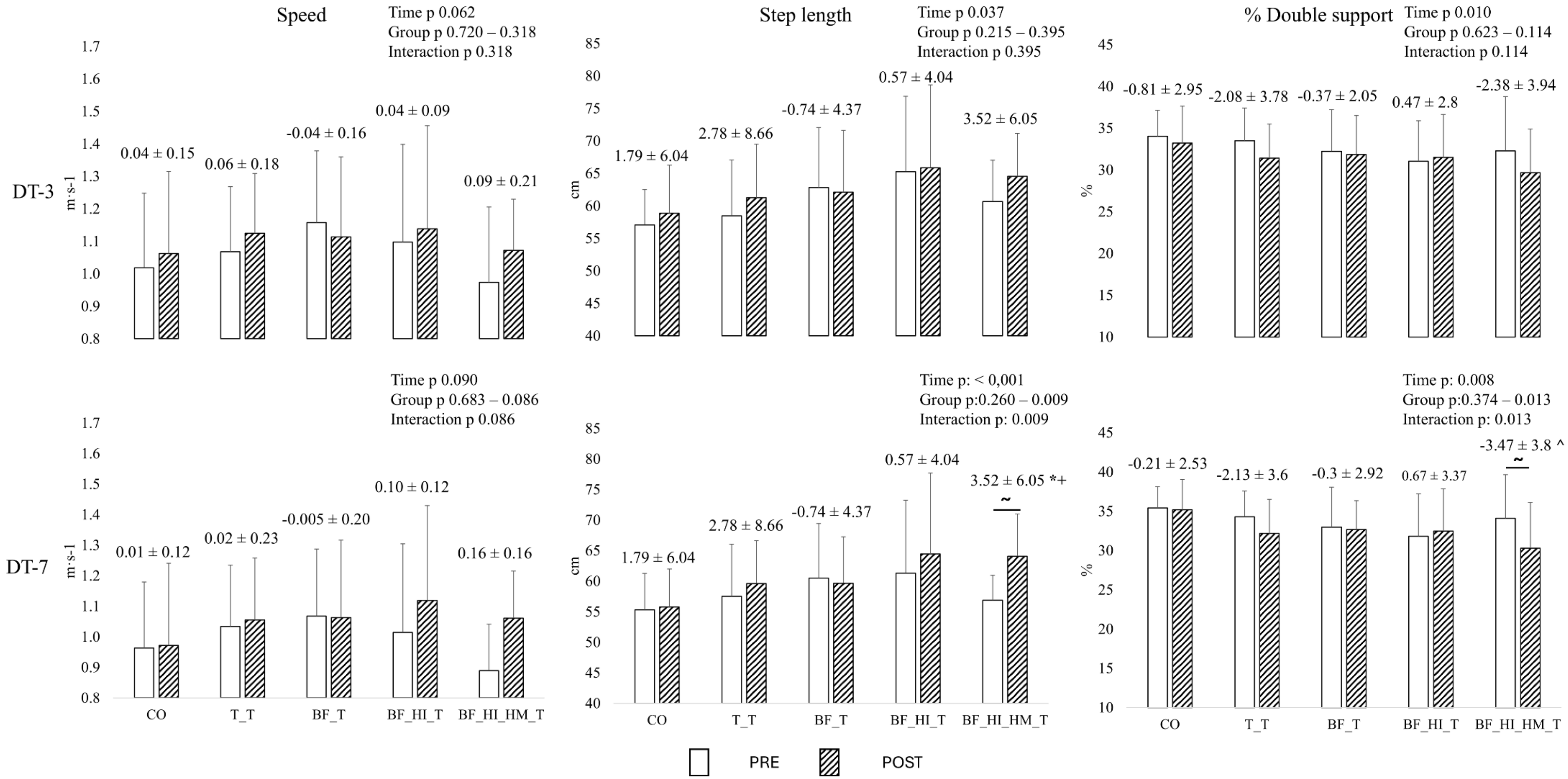

Figure 3 Panel B. For each group (C, T_T, BF_T, BF_HI_T, BF_HI_HM_T), PRE and POST values and within-group change (Δ) are reported under Speed ($m·s^{-1}$), Step length (cm), and Double support (DS, % gait cycle) in dual task conditions (DT-1; DT-7).
*Symbols indicate significant comparisons: ~ = PRE-POST within-group difference; * vs CO; # vs T_T; + vs BF_T; ^ vs BF_HI_*

5.3.3 Spatiotemporal Cost

For speed cost at DT-7, the Time × Group interaction was significant ($p = 0.042$; $\Delta p = 0.014$). No post-hoc comparisons (within or between groups) reached significance after correction. Descriptively, larger reductions were observed in the higher-interference groups: BF_HI_T decreased from 29.63 ± 14.12% (PRE) to 23.11 ± 15.03% (POST) ($p = 0.710$), $\Delta = -6.52 \pm 7.55$; BF_HI_HM_T decreased from 34.01 ± 8.38% to 25.92 ± 9.52 ($p = 0.666$), $\Delta = -7.36 \pm 7.61$, while changes in C, T_T and BF_T were small.

For step-length cost at DT-7, the Time × Group interaction was significant ($p = 0.035$; $\Delta p = 0.006$). Post-hoc tests did not reveal significant pairwise differences. Descriptively, the largest reductions were again in BF_HI_T, from 15.85 ± 8.27% (PRE) to 12.66 ± 9.51 (POST) ($p = 1.000$); $\Delta = -3.19 \pm 3.92$, and BF_HI_HM_T, from 18.88 ± 6.52% (PRE) to 13.47 ± 6.43 (POST) ($p = 0.366$); $\Delta = -4.92 \pm 6.02$, with minimal changes in the other groups.

The double-support cost at DT-1 showed a main effect of the Group ($p = 0.047$; $\Delta p = 0.181$), but no post-hoc pairwise contrasts reached significance. Concerning group means: BF_HI_T had the greatest reduction, from 13.25 ± 13.16% (PRE) to 6.94 ± 9.12% (POST) ($p = 1.000$); $\Delta = -6.31 \pm 15.64$, while the other groups experienced slight increases.

All remaining cost outcomes showed no significant main effects or interactions.

5.3.4 Difficulty Index

For the DI, a significant main effect of Time was observed at DT-1 ($p < 0.001$). In this case, the BF_HI_T group demonstrated a reduction from 0.78 ± 0.20 to 0.63 ± 0.20 ($p = 0.011$); $\Delta = -0.15 \pm 0.14$. Post-hoc comparisons on the change scores (Δ= POST−PRE) indicated a larger reduction in DI for BF_HI_T compared to Control (BF_HI_T: PRE 0.78 ± 0.20 to POST 0.63 ± 0.20, $\Delta = -0.15 \pm 0.14$; Control: PRE 0.62 ± 0.11 to POST 0.57 ± 0.22, Δ= 0.01 ± 0.09; $p = 0.021$ EF= -0.548).

The Time × Group interaction was significant at DT-3 ($p = 0.016$) and DT-7 ($p = 0.008$). At DT-7, BF_HI_T decreased from 3.20 ± 1.99 to 1.51 ± 0.82, $p = 0.042$; $\Delta = -1.92 \pm 1.63$. Post-hoc tests showed a greater DI reduction in BF_HI_T compared with Control (BF_HI_T: PRE $\Delta = -1.92 \pm 1.63$; Control: $\Delta = 1.25 \pm 2.14$; $p = 0.021$, ES = -0.636). In the same condition, BF_HI_HM_T differed from BF_HI_T, displaying a smaller improvement (BF_HI_HM_T: $\Delta = 0.08 \pm 1.16$; $p = 0.019$, ES = 0.552).

(Mean ± *SD*) Δ POST - PRE		CO		T_T		BF_T		BF_HI_T		BF_HI_HM_T				
		PRE	**POST**	**PRE**	**POST**	**PRE**	**POST**	**PRE**	**POST**	**PRE**	**POST**	*Time*	*Gr.*	*X*
DI [a.u.]	DT-1	0.62 ± 0.11	0.57 ± 0.22	0.99 ± 1.12	0.7 ± 0.17	0.8 ± 0.2	0.74 ± 0.14	0.78 ± 0.2	0.63 ± 0.2 ~	0.72 ± 0.26	0.62 ± 0.14	< 0.001	0.324	0.070
		0.01 ± 0.09		-0.02 ± 0.13		-0.07 ± 0.18		-0.15 ± 0.14 *		-0.09 ± 0.2			0.043	
	DT-3	1.36 ± 0.51	1.63 ± 1.08	2.25 ± 1.23	1.88 ± 0.95	1.61 ± 0.61	1.39 ± 0.47	1.3 ± 0.52	1.5 ± 0.9	1.78 ± 1.3	1.16 ± 0.24	0.235	0.245	0.016
		0.36 ± 0.74		-0.28 ± 0.72		-0.22 ± 0.42		0.19 ± 0.58		-0.56 ± 1.08			0.013	
	DT-7	2.62 ± 0.97	3.18 ± 2.51	4.05 ± 2.76	4.04 ± 2.59	3.44 ± 2.98	3.18 ± 1.36	3.2 ± 1.99	1.51 ± 0.82 ~	2.24 ± 1.1	2.33 ± 1.16	0.373	0.062	0.008
		1.25 ± 2.14		-0.37 ± 2.51		-0.3 ± 2.77		-1.92 ± 1.63 *		0.08 ± 1.16 ^			0.003	
Speed [m·s^{-1}]	Slope	-0.14 ± 0.06	-0.12 ± 0.12	-0.08 ± 0.07	-0.07 ± 0.06	-0.11 ± 0.07	-0.10 ± 0.04	-0.16 ± 0.12	-0.23 ± 0.15 #	-0.24 ± 0.15 #	-0.20 ± 0.14	0.996	0.001	0.064
		0.01 ± 0.09		0.01 ± 0.08		0.01 ± 0.09		-0.08 ± 0.12		0.04 ± 0.13			0.059	
	Intercept	1.25 ± 0.18	1.26 ± 0.19	1.25 ± 0.17	1.26 ± 0.2	1.32 ± 0.2	1.31 ± 0.2	1.35 ± 0.24	1.43 ± 0.21	1.28 ± 0.21	1.35 ± 0.15	0.153	0.326	0.431
		0.01 ± 0.1		0.01 ± 0.23		-0.02 ± 0.15		0.08 ± 0.13		0.08 ± 0.21			0.483	
Step length [cm]	Slope	-3.93 ± 2.03	-3.86 ± 3.07	-2.46 ± 2.28	-2.61 ± 2.39	-3.11 ± 1.62	-3.25 ± 2.12	-4.29 ± 3.33	-6.11 ± 4.79	-7.36 ± 4.8 #	-5.39 ± 4.27	0.987	0.016	0.106
		0.07 ± 2.19		-0.15 ± 2.14		-0.14 ± 2.93		-1.82 ± 4.13		1.97 ± 4.08			0.096	
	Intercept	63.58 ± 5.95	64.74 ± 6.34	64.37 ± 7.69	66.91 ± 8.13	67.54 ± 9.22	67.25 ± 7.65	71.24 ± 9.81	73.21 ± 8.76	69.61 ± 6.67	71.94 ± 7.3	0.036	0.044	0.591
		1.16 ± 3.7		2.54 ± 6.87		-0.3 ± 4.74		1.96 ± 4.57		2.33 ± 6.46			0.687	
DS [%]	Slope	1.43 ± 1.1	1.41 ± 1.25	1.09 ± 1.01	1.02 ± 0.99	1.34 ± 1.24	1.35 ± 0.55	1.7 ± 1.85	2.98 ± 2.91	3.05 ± 2.49	2.53 ± 1.84	0.72	0.045	0.065
		-0.02 ± 1.27		-0.07 ± 0.97		0.01 ± 1.28		1.28 ± 2.15		-0.51 ± 2.05			0.073	
	Intercept	32.14 ± 2.76	31.63 ± 3.81	31.26 ± 4.09	29.24 ± 4.21	29.73 ± 5.23	29.19 ± 4.1	28.32 ± 3	27.96 ± 3.04	28.67 ± 4.67	26.31 ± 4.42	0.014	0.041	0.315
		-0.51 ± 2.51		-2.02 ± 4		-0.54 ± 2.4		-0.35 ± 3.17		-2.35 ± 3.47			0.373	
Cadence [steps·min^{-1}]	Slope	-6.47 ± 4.29	-6.51 ± 7.04	-3.31 ± 2.97	-2.95 ± 2.6	-4.22 ± 3.38	-3.29 ± 2.35	-7.11 ± 5.81	-10.72 ± 7.55 #	-11.46 ± 5.76	-8.51 ± 5.56	0.841	0.006	0.036
		-0.04 ± 5.38		0.36 ± 3.82		0.94 ± 3.77		-3.61 ± 5.16		2.95 ± 5.14 ^			0.017	
	Intercept	115.7 ± 10.65	117.44 ± 11.62	115.07 ± 9.86	115.02 ± 9.32	116.01 ± 6.45	113.49 ± 7.36	113.34 ± 9.48	116.72 ± 6.12	113.03 ± 12.78	113.11 ± 6.87	0.735	0.880	0.495
		1.74 ± 6.77		-0.06 ± 10.2		-2.52 ± 7.77		3.38 ± 6.1		0.08 ± 10.51			0.453	

*Table 4. DI and linear relations with gait parameters. For each group (Control, T_T, BF_T, BF_HI_T, BF_HI_HM_T), PRE and POST values (mean ± SD) and change (Δ = POST − PRE) are shown. The first block reports DI at DT-1, DT-3, and DT-7. Subsequent blocks report slope and intercept of the DI relations with Speed (m·s^{-1}), Step length (cm), Double support (% gait cycle), and Cadence (steps·min^{-1}). Statistics symbols: ~ within-group PRE-POST difference; * vs C; # vs T_T; + vs BF_T; ^ vs BF_H*

5.3.5 Linear relations between DI and spatiotemporal parameters

Across models, the Group main effect was significant for the speed-DI slope ($p = 0.001$), the step length-DI slope ($p = 0.016$), and % DS-DI slope ($p = 0.045$). For the intercepts, a Time main effect appeared for step length ($p = 0.036$), and both Time ($p = 0.014$) and Group ($p = 0.041$) were significant for double support; all other omnibus tests on slopes/intercepts were not significant. Except where noted below, post-hoc contrasts did not survive correction.

Within this framework, three pairwise contrasts were identified as significant. Firstly, at POST, the speed slope was more negative in BF_HI_T than in T_T (−0.23 ± 0.15 vs −0.07 ± 0.06; $p = 0.007$). Secondly, at PRE, the speed slope in BF_HI_HM_T was more negative than in T_T (−0.24 ± 0.15 vs −0.08 ± 0.07; $p = 0.034$). Thirdly, at PRE, the step-length slope in BF_HI_HM_T was more negative than in T_T (−7.36 ± 1.80 vs −2.46 ± 2.28; $p = 0.036$). Post-hoc comparisons, adjusted for multiple testing via the Holm method, were not significant, except where explicitly indicated.

Only significant tests are reported; all other effects were non-significant. Symbols used in the figures: ~ = within-group PRE-POST difference; * vs Control; # vs T_T; + vs BF_T; ^ vs BF_HI_T (*Table 4*).

5.4 Discussion

This study assessed the effectiveness of five four-week aerobic training protocols for individuals with PD, focusing specifically on treadmill walking with BF delivered at low and high contextual interference levels, as well as a protocol combining high interference with increased metabolic intensity. Beyond testing a single training mode, the design intentionally examined gait adaptations across both single-task and DT conditions, capturing training effects under increasing levels of attentional demand. An additional strength is the individualisation of training intensity (progressions anchored on each participant's self-selected speed and disease stage) and feedback targets (step length and cadence), which were tailored to normative, height- and age-adjusted references. This approach ensures that the spatiotemporal goals are clinically meaningful for each participant. Finally, all interventions adhered to standardised protocols with predefined progression session structures, dosages, and staged increases in task difficulty, thereby enhancing fidelity and reproducibility. This also allows for precise between-group comparisons of how feedback complexity and, in one intervention, added metabolic load influence gait outcomes. Against this standardised and individualised training framework, we first assessed primary spatiotemporal outcomes.

Overall, gait speed showed no differential change: Time × Group interactions were not significant across all conditions (ST, DT-1, DT-3, DT-7; all $p > 0.05$), and main effects of Time and Group also did not reach significance (ST $p = 0.060$; DT-1 $p = 0.142$; DT-3 $p = 0.062$; DT-7 $p = 0.090$; all Group

$p \geq 0.350$). This pattern indicates that, over a four-week period, walking speed remained largely stable regardless of intervention, consistent with previous research demonstrating that speed is a higher-level composite that can be less sensitive in the short term than more specific spatiotemporal measures such as step length and double support (Mehrholz et al., 2015; Ni et al., 2018).

By contrast, step length generally tended to improve, with the main effect of Time being significant in ST, DT-3, and DT-7. Context specificity appeared in DT-7 (Time × Group $p = 0.009$). In this most challenging condition, BF_HI_HM_T showed a significant within-group increase (PRE 56.9 ± 4.08 → POST 64.09 ± 6.91; $p = 0.003$) and a larger Δ than CO (Δ 6.54 ± 6.29 vs 0.47 ± 4.27; $p = 0.036$; ES = −1.285), and BF_T (Δ 6.54 ± 6.29 vs −0.84 ± 5.04; $p = 0.007$; ES = 1.53). Overall, the effects of time, the significant interaction in DT-7, and post-hoc contrasts with large effect sizes suggest a clinically meaningful improvement in step amplitude under high attentional demands within the BF groups, consistent with evidence that real-time, task-relevant feedback can enhance motor learning and strengthen spatiotemporal adaptations in PD (Baudendistel et al., 2024; McMaster et al., 2022; Ni et al., 2018).

For double support, Time was significant in all conditions (ST $p = 0.002$; DT-1 $p = 0.025$; DT-3 $p = 0.010$; DT-7 $p = 0.008$), with Time × Group interactions in ST ($p = 0.011$) and DT-7 ($p = 0.013$). In ST, BF_HI_HM_T showed a lower POST value than CO (25.58 ± 4.28 vs 31.66 ± 3.81; $p = 0.016$), indicating greater step stability (meaning a shorter proportion of the gait cycle was spent in double support). In DT-7, BF_HI_HM_T showed a significant reduction within the group (PRE 34.11 ± 5.53 → POST 30.29 ± 5.82; $p = 0.022$) and a Δ that differed from BF_HI_T (Δ −2.65 ± 2.89 vs 0.95 ± 3.01; $p = 0.019$; ES = 1.37). The convergence of global effects with post-hoc contrasts and sizeable effect sizes indicates a notable enhancement in step stability during high-difficulty DT conditions in the BF groups, aligning with previous reports that targeted feedback can improve temporal control of the gait cycle in PD (Ni et al., 2018). However, the interpretation of the pronounced benefits observed in the BF_HI_HM_T group must be carefully contextualised by their baseline characteristics. As noted in the results, this group exhibited significantly lower motor severity (UPDRS-III) and higher baseline aerobic capacity (peak power output) than the other groups. These baseline advantages may have provided these participants with greater functional and cognitive reserve, enabling them to better tolerate and adapt to the dual stress of high-interference feedback and increased metabolic load. Consequently, while the combined intervention proved highly effective for this subset of patients, their milder initial impairment likely facilitated their engagement with the demanding tasks. It remains uncertain whether individuals with more advanced motor or cognitive impairment would achieve similar adaptations or if they would find the combined load too overwhelming, as suggested by the higher perceived exertion (RPE) and the single dropout in this group. Because mean changes might hide difficulty-dependent behaviour, we then evaluated cost-based and composite difficulty metrics.

For spatiotemporal costs (speed and step-length costs), DT-7 exhibited significant Time × Group interactions (speed cost: $p = 0.042$; step-length cost: $p = 0.035$). However, none of the pairwise post-hoc comparisons remained significant after the Holm correction. Descriptively, the largest pre-post reductions were observed in the high-interference intervention, surpassing the changes in the comparator groups (CO, T_T, BF_T): speed cost Δ was −6.52 ± 7.55% in BF_HI_T and −7.36 ± 7.61% in BF_HI_HM_T; step-length cost Δ was −3.19 ± 3.92% in BF_HI_T and −4.92 ± 6.02% in BF_HI_HM_T. Given the short, four-week duration and individual variability, these findings should be regarded as exploratory indications of improvement relative to the control, pending confirmation in longer trials with increased power.

Regarding the DI, this metric offers a combined assessment of motor-cognitive demand during DT walking; therefore, a decrease in DI can be seen as a general improvement in managing the concurrent cognitive task while maintaining gait. This aligns with previous research showing that DT paradigms increase gait deficits in PD and are often summarised with cost- or composite index beyond raw speed (Kelly et al., 2012; Vitorio et al., 2021). Furthermore, the reliability of quantitative gait assessment under single- and DT conditions has been established in older populations, supporting the use of such composite measures to capture DT burden. (Montero-Odasso et al., 2009). In PD specifically, dual tasking influences spatiotemporal parameters, particularly stride length and speed, more than in healthy controls, emphasising the construct validity of difficulty/cost index for measuring motor-cognitive load (Salazar et al., 2017).

Where post-hoc contrasts confirmed differences, the magnitude of change supports a functional interpretation. In DT-1, the Δ for BF_HI_T was more favourable than Control (BF_HI_T: −0.15 ± 0.14 vs Control: +0.01 ± 0.09), with ES 1.29 (very large). In DT-7, the advantage of BF_HI_T over Control was even greater (BF_HI_T: −1.92 ± 1.63 vs Control: +1.25 ± 2.14), ES 1.64 (very large). Conversely, BF_HI_HM_T showed a smaller improvement than BF_HI_T in DT-7 (BF_HI_HM_T: +0.08 ± 1.16 vs BF_HI_T: −1.92 ± 1.63), ES 1.32 (large). In addition to evidence that real-time, task-relevant BF can alter PD gait characteristics and improve specific spatiotemporal features, these effects suggest a significant reduction in DT difficulty under high-interference BF (Baudendistel et al., 2024; McMaster et al., 2022; Ni et al., 2018). Simultaneously increasing the metabolic load (incline, weighted vest) did not seem to further decrease overall DT difficulty, as measured by DI, beyond the benefit seen with high interference alone. Building on the DI framework described above, we explored linear models to characterise sensitivity to difficulty and baseline levels.

The linear models connecting the DI to speed, step length, and percentage of double support (%DS) quantify how spatiotemporal control varies with DT difficulty. In this model, the slope indicates how sensitive a gait parameter is to changes in DI (how much the parameter varies for a unit change in DI), while the intercept shows the expected value of the parameter when DI = 0 (i.e., the baseline level when DT difficulty is minimal).

At the omnibus level, group effects on slopes for speed-DI, step length-DI, and %DS-DI indicate that the strength of the relationship between DI and each parameter varied across interventions. By "stronger linear combining", we mean more negative slopes, meaning that increases in DI (i.e., more difficulty) are linked to larger decreases in speed or step length (or larger increases in %DS), and vice versa, decreases in DI are linked to greater gains in these parameters. In specific pairwise tests, this appeared as steeper (more negative) speed-DI slopes in BF_HI_T versus T_T at POST (-0.23 ± 0.15 versus -0.07 ± 0.06; $p = 0.007$), and in BF_HI_HM_T versus T_T at PRE (-0.24 ± 0.15 versus -0.08 ± 0.07; $p = 0.034$), as well as a steeper step length-DI slope in BF_HI_HM_T versus T_T at PRE (-7.36 ± 1.80 versus -2.46 ± 2.28; $p = 0.036$).

For the intercepts, a Time effect on step length and %DS (plus an additional Group effect for %DS) indicates training-related changes in the DI=0 baseline of these parameters. Functionally, a higher intercept for step length after training suggests a larger step amplitude even at minimal DT difficulty, while a lower intercept for %DS indicates less time in double support, meaning greater step stability at DI=0. These shifts are most consistent with improvements in groups exhibiting larger post-training step length and lower percentage of double support (DS), particularly in the high-interference, higher metabolic intensity group, where increases in step length and decreases in DS reflect a better baseline gait pattern (at DI=0) alongside any changes in slope.

Whereas spatiotemporal adaptations developed within four weeks, cardiorespiratory outcomes showed no corresponding change. On the cardiorespiratory side, no significant differences were observed in HR_max, $\dot{V}O_2max$, GET, or RCP during the four-week intervention, although PPO differed between groups at baseline and over time, reflecting the initial differences. Conversely, randomised trials with longer durations have reported improvements in aerobic fitness. For example, the Park-in-Shape trial demonstrated increases in $\dot{V}O_2max$ after six months of home-based aerobic cycling with gamified feedback and remote supervision, compared to a stretching control (three sessions per week) (Johansson et al., 2022). Similarly, a randomised clinical trial comparing treadmill versus stretching/resistance over three months (three sessions per week) found a 7-8% increase in peak $\dot{V}O_2$ in both treadmill groups compared to the non-aerobic comparator (Shulman et al., 2013). These data position the current four-week null results for cardiorespiratory outcomes within evidence that extended training periods (≥3-6 months) can produce measurable improvements in aerobic capacity in PD.

Adherence to the training protocol remained consistently high across all groups, with no significant differences between groups in the number of completed sessions ($p = 0.497$); on average, participants attended more than 11 out of 12 sessions per group. The mean treadmill walking speed during training also did not differ between treadmill-based conditions ($p = 0.604$), with average speeds ranging from 1.3 to 1.4 $m \cdot s^{-1}$. Notably, only one participant discontinued the study, and this occurred in the high-interference, higher-metabolic group due to difficulties in performing the cognitively and metabolically more demanding training.

Regarding perceived exertion (RPE), a significant overall group effect was detected ($p = 0.033$), although post-hoc comparisons did not reach the corrected significance threshold. Nonetheless, the nearly significant contrasts and large effect sizes suggest a consistent pattern: BF_HI_HM_T reported higher RPE than T_T ($p = 0.058$, ES = 1.10) and BF_T ($p = 0.051$, ES = −1.15). This adherence and RPE data support feasibility, while the higher perceived effort in BF_HI_HM_T highlights the need for careful titration when extending to more advanced PD.

This study has several linked strengths. First, we used a standardised yet personalised protocol, where treadmill speed and BF targets (step length, cadence) followed predefined progressions but were tailored to each participant using normative, height- and age-adjusted references and stratified by H & Y stage; this ensured that training goals were both comparable across participants and clinically meaningful for each individual (Moe-Nilssen and Helbostad, 2020). Second, we implemented multiple intervention groups with comparable training volumes but systematic variation in feedback modality (low- versus high-interference) and metabolic intensity. This design enabled direct comparisons between feedback strategies while keeping the training volume constant, thereby isolating the specific contribution of attentional interference and distinguishing it from dose effects. Third, we demonstrated the practical feasibility of delivering real-time gait BF using an OptoGait-based setup mounted on commercial treadmills, supporting the reproducibility and scalability of the protocol beyond research laboratories and facilitating translational uptake in rehabilitation services that already use standard clinical equipment.

5.5 Limitations and Future Directions

This study has several limitations. The duration was short (4 weeks), which may have limited the emergence and consolidation of cardiorespiratory and higher-order gait adaptations. Most notably, baseline imbalances were present for key clinical and performance indicators (e.g., significantly lower

UPDRS-III and higher PPO in the BF_HI_HM_T group). Although our linear mixed-effects models statistically accounted for these discrepancies by isolating the interaction with time, these imbalances remain a clinical confounding factor. Patients with milder symptoms inherently possess greater neuroplasticity and motor learning capacity, which limits our ability to definitively attribute the superior outcomes of the BF_HI_HM_T group solely to the training protocol. Furthermore, drop-outs occurred in the high metabolic-intensity group, potentially reducing statistical power and causing differential attrition. Instrumentation heterogeneity (three treadmill systems) may have increased measurement variability, and assessments were performed in the ON phase without strict synchronisation to dopaminergic dosing, affecting pharmacological timing variability. The lack of follow-up prevents the estimation of retention and transfer to overground walking. Additionally, DT testing involved cognitive demands that were not fully aligned with those emphasised during training, which limits task-specific inferences. Finally, the single-task assessment used self-selected overground speed rather than treadmill-determined speed, which may have created a mismatch between the testing and intervention contexts.

Future work should prioritise longer interventions (≥ 8-12 weeks) with post-intervention follow-up to measure retention and real-world transfer. Trials should clarify dose-response relationships for the attentional-interference component of BF and for metabolic loading, ideally customised to physiological thresholds (e.g RCP). To improve ecological validity and task specificity, we suggest testing with single and dual tasks on instrumented or mechanical treadmills that allow participant-specific speed selection, thereby matching the measurement environment to the training setting. Furthermore, including field measures of functional capacity, especially the 6-Minute Walk Test, would complement treadmill assessments by gauging endurance and walking performance during sustained, clinically relevant effort. These suggestions align with the growing use of treadmill-based real-time BF to target spatiotemporal regulation in PD (Baudendistel et al., 2024).

5.6 Conclusions

In this study of individuals with PD, a four-week aerobic training programme with treadmill and BF variants demonstrated that, despite the lack of systematic change in gait speed, there were significant improvements in step length and double support, especially under the most demanding DT condition (DT-7) and in the high-interference BF training. Specifically, these gains also appeared when high interference was combined with a higher metabolic load, notably a within-group increase in step length in DT-7 and reductions in double support in ST and DT-7 for the BF_HI_HM_T group. Post-hoc contrasts, supported by large effect sizes, show greater step amplitude and improved temporal stability of the gait cycle, both key factors for functional walking under DT demands. Consistently, the DI, seen as a combined measure of motor-cognitive demand, decreased more significantly with high-interference BF, indicating better concurrent management of the cognitive task and gait control. The slope/intercept analyses linking DI to spatiotemporal parameters also suggest increased sensitivity to difficulty (steeper slopes) and improved DI = 0 baselines (longer step length, lower percentage of double support), which aligns with a better underlying gait profile after training.

On the cardiorespiratory side, no significant changes were observed over four weeks in HRmax, $\dot{V}O_2max$, GET, or RCP, whereas between-group differences in PPO were consistent with baseline disparities. Adherence was high and consistent across groups (>11/12 sessions), with 2 dropouts in the most demanding group. Perceived exertion (RPE) tended to be higher in the high-interference, higher-metabolic intensity protocol, a clinically relevant consideration for translation to more advanced disease stages.

Taken together, these findings support the feasibility and clinical usefulness of protocols that are standardised yet tailored, with predefined progression speeds and step length/cadence targets based on normative references and H & Y stages, and that can be implemented on commercial treadmills equipped with OptoGait. The evidence suggests that high-interference BF, specifically designed to target spatiotemporal regulation, can yield significant improvements in DT walking over a short period. Future research with longer durations (≥8-12 weeks) and follow-up assessments is necessary to understand dose-response relationships for the attentional interference aspect and personalised metabolic loading, to determine whether these protocols can also enhance aerobic capacity, and to assess the retention and transfer of benefits to everyday overground walking.

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CHAPTER 6: General Discussion

6.1 Final Remarks

The primary aim of this doctoral research was to develop evidence-based strategies for evaluating and rehabilitating gait impairments in PD, focusing on how cognitive load and targeted treadmill training can influence spatiotemporal gait control. While the first two studies established the methodological and mechanistic foundations of motor-cognitive interference in gait, by developing a graded DT protocol and characterising its effects in early-stage PD, the final study provided the translational core of the thesis. The study investigated whether treadmill-based BF interventions, especially those involving high contextual interference and increased metabolic intensity, could significantly improve gait performance under single- and DT conditions. This intervention aspect is therefore the main contribution of the thesis and shapes the interpretation of the previous findings within a rehabilitation-focused perspective.

The development of the graded DT protocol served as both a conceptual and methodological foundation for the training work. By demonstrating a strong and scalable link between increasing cognitive load and declines in gait speed, step length, cadence, and double support, the initial study provided a controlled framework for measuring the attentional vulnerability of gait. These insights were crucial for understanding which aspects of locomotor control could benefit from targeted rehabilitation, especially those related to automaticity and the ability to stabilise gait when attentional resources are limited. Similarly, applying this graded paradigm to individuals with early-stage PD confirmed that people with PD not only demonstrate impaired gait automaticity under single-task conditions but also experience disproportionate DT costs and tend to fall below safety thresholds under higher cognitive loads. In this context, the test provided a window into the specific motor-cognitive deficits that rehabilitation should aim to address.

However, the most clinically relevant question, whether targeted practice can reduce these vulnerabilities, was addressed in the third study. Here, treadmill training enhanced with real-time BF offered a high level of mechanistic relevance for motor-cognitive integration: participants were required to monitor and adjust gait parameters dynamically, simulating the type of divided attention required in complex environments. However, it must be acknowledged that direct transfer to daily life mobility or ecological overground settings was not assessed in this study. The interventions varied systematically in complexity and metabolic load, allowing the identification of the conditions under which gait adaptations are most likely to occur.

Over four weeks, the most consistent and significant improvements were seen in the high-interference BF groups, especially under the most challenging DT condition (DT-7). These participants showed longer step length and less double support, both of which indicate better gait automaticity and stability, despite no significant change in overall gait speed. Notably, these changes occurred precisely where gait was previously identified as most vulnerable: during high cognitive load. This targeted improvement reinforces the interpretation that high-interference BF enhances not only motor recalibration but also the robustness of gait control when attentional resources are limited. Simultaneously, the DI decreased significantly in the high-interference group, indicating a more efficient management of concurrent motor and cognitive tasks. Linear modelling further revealed more advantageous post-training baselines (e.g., higher intercepts for step length and lower intercepts for double support), aligning with improved underlying gait patterns that were unaffected by the difficulty level.

The combined high-interference/high-metabolic intensity protocol yielded similar or even greater improvements, especially in the most challenging conditions, although perceived exertion was higher and some participants raised feasibility concerns. This indicates that metabolic intensity may enhance gait-related adaptations when suitably individualised, but it must be balanced with tolerability, particularly in later stages of disease. Notably, aerobic capacity did not improve over the short, four-week training period, a finding consistent with previous evidence that changes in cardiorespiratory fitness usually require longer interventions.

Together, these results indicate a coherent interpretation: training methods that combine attentional challenges with real-time, task-relevant sensory feedback have the greatest potential to enhance the spatiotemporal features of gait, which are most affected by cognitive load in PD. The graded DT findings from earlier studies identified where the deficits occur; the training study demonstrated how targeted, high-interference practice can alter those very parameters.

This integrated perspective supports a modern understanding of PD gait: not just a motor deficit but a deficit of automaticity, where executive and attentional resources must compensate for impaired basal ganglia function. Rehabilitation, therefore, cannot rely solely on repetitive motor practice but must include attentional modulation, variability in context, and environments rich in feedback. By demonstrating that such training enhances gait under cognitively demanding conditions, this thesis supports the development of interventions explicitly designed to operate at the intersection of attention and locomotor control.

In summary, the training-focused findings represent the translational application of the methodological and mechanistic work, showing that targeted BF, especially in settings with high contextual interference, can greatly enhance gait performance in PD, particularly during DT walking where it is most at risk.

6.2 Summary of Findings

6.2.1 Dual-Task Assessment in Healthy Adults and Parkinson's Disease

Across the first two studies, a graded serial-subtraction DT protocol (-1, -3, -7) was developed and validated as a sensitive tool to characterise motor-cognitive interference in gait. In healthy older adults, increasing cognitive load led to predictable reductions in gait speed, step length, and cadence, accompanied by a rise in double support. DTC increased proportionally with task difficulty, confirming a clear dose-response relationship.

The introduction of the DI, based on the number of correct responses per second, enabled continuous rather than categorical measurement of cognitive load. DI closely correlated with spatiotemporal decline, capturing individual load-response patterns and providing a unified metric that reduces variability between individuals caused by differences in numeracy or task strategy.

When applied to early-stage PD, the graded protocol uncovered impaired gait automaticity even under single-task conditions. During a DT load, both PD and control participants exhibited graded deterioration; however, PD participants walked more slowly, took shorter steps, and had greater double support across all conditions. DTC were consistently higher in PD, and many individuals fell below the clinically relevant threshold of 1.0 $m \cdot s^{-1}$ under increased loads. Importantly, DI-gait slopes were broadly similar across groups, indicating comparable sensitivity to cognitive load despite worse baseline levels in PD. These findings support the utility of the graded DT test and DI as sensitive markers of motor-cognitive impairment and early functional decline in PD.

6.2.2 Training Outcomes

The third study applied these insights to a rehabilitation setting. Over four weeks of treadmill training, BF interventions, particularly those involving high contextual interference, led to significant improvements in spatiotemporal gait control. While gait speed remained mostly unchanged, high-interference training caused notable increases in step length and reductions in double support during the most demanding DT condition (DT-7). Protocols combining high interference with high metabolic intensity showed similar or even greater gains but required higher perceived effort. These improvements were accompanied by decreases in the DI, indicating better management of motor-cognitive load. Linear modelling revealed positive shifts in intercepts for step length and double support, suggesting improved baseline gait patterns regardless of task difficulty. No changes in aerobic fitness were observed over the four weeks, aligning with evidence that longer interventions are needed to alter cardiorespiratory parameters. Overall, the training confirmed that interventions integrating attentional challenge, contextual variability, and real-time feedback can enhance gait automaticity and stability in PD under laboratory DT conditions. While these spatiotemporal improvements are theoretically linked to better dynamic balance, actual fall-risk reduction and transfer to overground, real-world mobility were not directly measured. Therefore, whether these laboratory-based gains translate into fewer falls in daily life remains a key hypothesis to be explicitly tested in future prospective trials.

Moving beyond the direct conclusions supported by the presented data, we hypothesise that from an implementation perspective, the training methodology developed in this thesis could be feasible and scalable beyond the experimental setting. The standardised yet individually tailored protocol might be delivered using commercially available treadmills equipped with the OptoGait system used in this work, which effectively transforms the device into an "intelligent" treadmill capable of providing task-relevant, real-time feedback on spatiotemporal gait parameters. In specialised neurorehabilitation centres, such a setup could be incorporated into structured exercise-therapy pathways for people with PD, enabling therapists to adjust cognitive load, BF complexity, and metabolic intensity based on disease stage and individual goals. Similarly, health-focused gyms or community-based "exercise for PD" programmes could utilise the same system to provide supervised, disease-specific training sessions in a more accessible setting. Importantly, using commercially available treadmills and modular optoelectronic systems can significantly lower the financial barrier compared to dedicated, highly specialised gait-training treadmills, whose purchase and maintenance costs are often unaffordable for community facilities and domestic settings. Ultimately, the widespread availability of domestic treadmills indicates the potential to expand this approach to home-based practice. However, this shift would require further development of hardware and software to provide user-friendly, pre-programmed training modules and to guarantee safety without direct supervision. Future versions of the system should include intuitive interfaces, remote monitoring options, reliable safety features, and affordable design to support independent use at home. This would allow optical BF treadmill training to move from an experimental concept to a practical, scalable, and cost-effective exercise therapy solution in clinical, community, and domestic settings.

6.3 Experimental Considerations and Future Directions

Several methodological considerations arise from this research programme. While the graded DT protocol and the DI show robustness and clinical usefulness, future studies would benefit from including longer walking periods or wearable sensor-based evaluations to better capture gait variability and smoothness.

In the training context, the four-week duration, although sufficient to elicit changes in step length and double support, was probably too short to induce significant adaptations in aerobic capacity. Future interventions should therefore focus on longer protocols ($\geq$ 8-12 weeks) with both linear and non-linear progressions in training load. Non-linear progressions, characterised by fluctuating cognitive demands, alternating high- and low-interference blocks, variable BF frequency, or intermittent perturbations, may enhance motor learning by increasing contextual diversity and promoting flexible, generalisable gait control. These progressions could also include non-linear metabolic loading patterns (e.g., interval-based incline changes or variable weighted-vest intensity) to better target physiological adaptation.

To ensure ecological validity, future research should include follow-up assessments and functional tests, such as the 6-Minute Walk Test, alongside treadmill- and overground DT evaluations. Lastly, aligning training tasks more closely with the DT assessments by integrating similar cognitive demands or dynamic task-switching within gait practice may enhance transfer and improve the specificity of adaptations.

Alongside these protocol improvements, future research should also focus on the technological, organisational, and economic aspects of implementing the training platform. As previously mentioned, combining a commercial treadmill with an OptoGait system provides a relatively straightforward way to convert standard devices into “intelligent” treadmills capable of offering personalised, feedback-driven exercise therapy. However, wider adoption will depend on user-friendly software for setting up and scheduling personalised training programmes, smooth integration with clinical workflows and tele-monitoring systems, and strong safety features that support both supervised and, in the longer term, semi-supervised or home-based use. From an economic perspective, repurposing existing commercial treadmills with add-on OptoGait systems may substantially reduce initial costs compared to buying specialised instrumented or robotic gait-training treadmills. This could enhance feasibility in resource-limited rehabilitation services, health-focused gyms, and possibly in selected home settings. Creating standardised configuration templates for various clinical profiles and settings (such as specialised rehabilitation centres, health-focused gyms, and domestic environments) would further support integration into routine practice and enable future multicentre trials not only to assess the generalisability and long-term effects of intelligent treadmill training in PD, but also to appraise its cost-effectiveness compared to more costly, specialised gait-training technologies.