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Improved Dual-Task Interference in Parkinson’s Disease following Deep Brain Stimulation / Ghislieri, M., Locoratolo, L., Sciscenti, F., Knaflitz, M., Rizzi, L., Armocida, D., Lanotte, M., Agostini, V.. - In: IEEE TRANSACTIONS ON NEURAL SYSTEMS AND REHABILITATION ENGINEERING. - ISSN 1534-4320. - ELETTRONICO. - 33:(2025), pp. 4624-4635. [10.1109/tnsre.2025.3632674]

Availability:
This version is available at: 11583/3005172 since: 2025-12-01T13:45:02Z

Publisher:
IEEE

Published
DOI:10.1109/tnsre.2025.3632674

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13 November 2026

Improved Dual-Task Interference in Parkinson's Disease Following Deep Brain Stimulation

Marco Ghislieri, Lorenzo Locoratolo, Fabrizio Sciscenti, Marco Knaflitz, Laura Rizzi, Daniele Armocida, Michele Lanotte, and Valentina Agostini

***Abstract*— This work aims to evaluate how Deep Brain Stimulation (DBS) impacts the motor-cognitive dual-task performance of Parkinson's Disease (PD) patients. We analyzed the muscle synergies of 27 PD patients at T_0 (pre-surgery), T_1(3 months post-surgery), and T_2 (12 months post-surgery), compared to a control group of 30 age-matched individuals, during a walking task and a motor-cognitive dual task (walking while performing a phonemic fluency task). To evaluate dual-task interference, the Dual Task Effect (DTE) of both motor and cognitive metrics was analyzed. Our findings demonstrate that DBS significantly enhances dual-task capacity, with PD patients transitioning from a detrimental "posture-second" strategy at T_0 to a more efficient attentional allocation pattern post-surgery. More specifically, the average motor metric $mDTE_{FWHM}$ (DTE of the Full-Width-at-Half-Maximum of the muscle synergy activation coefficients) of PD patients changed from -12.5 ± 11.5 % (T_0 to -3.7 ± 10.2 % (T_1 and -4.5 ± 8 % (T_2, becoming not different from that of controls (-1.1 ± 12.7 %). On the other hand, the PD cognitive DTE (cDTE) at T_0 was -12.4 ± 23.2 %, not significantly different from that of controls (-23.0 ± 21.0 %), and remained unchanged 1 year after the DBS implant (T_2: -11.2 ± 25.9). The reduced impact of cognitive loading on motor function without compromising cognitive performance suggests enhanced attentional resource management of PD patients after DBS that may translate to improved dynamic balance and reduced fall risk in daily activities.**

***Index Terms*— DBS, DTE, dual-task walking, muscle synergies, PD.**

I. Introduction

EVERYDAY-LIFE activities rarely involve a single, isolated task; rather, they typically require the simultaneous execution of cognitive and motor tasks, such as talking while walking or texting while navigating a busy environment [1], [2]. Given the central role of walking in patients' daily life and overall quality of life, dual-task walking has become a valuable paradigm for assessing motor–cognitive impairments in various neurological disorders [3], [4], [5]. A key aspect of the dual-task paradigm is the concept of the Dual-Task Effect (DTE), which quantifies the relative change in motor (or cognitive) performance when tasks are performed concurrently compared to in isolation. By evaluating the DTE, researchers can objectively measure the dual-task interference which is the extent to which dual-tasking impacts motor (or cognitive) performance in everyday-life contexts. To account for these performance changes, the "capacity sharing" model has been proposed and is now widely recognized in the literature [6]. This model states that a limited pool of attentional resources must be divided among concurrent tasks. When the combined demands of these tasks exceed available resources, a trade-off becomes necessary: individuals may prioritize motor performance (e.g., maintaining balance and preventing falls) at the expense of cognitive function or cognitive performance (e.g., focusing on complex mental calculations) at the expense of motor function. This trade-off is especially pronounced in neurological populations, such as individuals with Parkinson's Disease (PD), who typically require additional attentional resources to compensate for motor symptoms [5]. As a result, fewer attentional resources are available for the simultaneous execution of secondary tasks, making individuals with neurological deficits more susceptible to dual-task-related impairments [7], [8].

While the detrimental effects of dual-tasking on motor-cognitive performance in people suffering from PD are documented in the literature [5], [7], [8], the impact

Received 5 August 2025; revised 8 November 2025; accepted 11 November 2025. Date of publication 13 November 2025; date of current version 25 November 2025. This work was supported by European Union—Next Generation EU within the PRIN 2022 Program (D.D. 104—2 February 2022 Ministero dell'Università e della Ricerca) through the PD-DBS Project "Effect of Bilateral Subthalamic Nucleus Deep Brain Stimulation on Gait Analysis and Muscle Synergy Patterns of Patients Affected by Parkinson's Disease During Dual-Task Walking", Protocol Nº under Grant 2022KWSJJT. *(Marco Ghislieri and Lorenzo Locoratolo are co-first authors.) (Corresponding author: Michele Lanotte.)*

This work involved human subjects or animals in its research. Approval of all ethical and experimental procedures and protocols was granted by the Ethics Committee of A.O.U. Città della Salute e della Scienza di Torino—A.O. Ordine Mauriziano—A.S.L. "Città di Torino" under Application No. 0092029.

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This article has supplementary downloadable material available at https://doi.org/10.1109/TNSRE.2025.3632674, provided by the authors.

Digital Object Identifier 10.1109/TNSRE.2025.3632674

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of therapeutic interventions (both pharmacological and non-pharmacological treatments) on motor-cognitive performance under dual-task conditions remains largely unexplored. For example, despite high-frequency Deep Brain Stimulation (DBS) of the subthalamic nucleus being one of the most widely used non-pharmacological treatments for PD [9], its effects on motor-cognitive interference during dual-tasking is poorly understood [10]. Additional studies are therefore essential to inform clinical practice and improve patient outcomes, as well as to address the main limitation of current research, which rarely analyzes reciprocal dual-task effects (i.e., the assessment of cognitive and motor performance in both single-task and dual-task conditions). This limitation significantly limits our understanding of the effects of Parkinson's disease or stimulation treatment on dual-task performance [11].

In studies investigating motor-cognitive dual-task effects during gait, cognitive performance is typically assessed using standardized tasks such as phonemic verbal fluency, with the number of words produced serving as a quantitative measure. Motor performance, instead, is generally evaluated by measuring gait speed. Complementing traditional motor performance outputs, the muscle synergy framework has been introduced as a non-invasive means of probing the neuromuscular control strategies employed by the Central Nervous System (CNS) [12], [13]. Muscle synergy theory suggests that the CNS simplifies the coordination of numerous muscles by activating them in groups or "synergies", effectively reducing the complexity and degrees of freedom inherent in motor control. By extracting these synergies from surface electromyographic (EMG) signals, researchers can quantitatively and non-invasively characterize alterations in motor control patterns. This approach has proven valuable not only in distinguishing individuals with neurological impairments (including PD) from healthy individuals [14], [15], but also in capturing and describing motor improvements observed in PD patients following DBS implantation [15].

Despite the growing interest in both dual-task walking and muscle synergy theory, only a few studies have explored dual-task effects during walking in healthy subjects by using these concepts together [16], [17]. Nevertheless, the integration of dual-task paradigms with muscle synergy analysis offers a robust, multimodal framework to investigate how increased cognitive load impacts motor control of patients suffering from PD before and after DBS neurosurgery. This framework could not only enhance our understanding of attentional resource allocation and neuromuscular strategy adaptations in PD, but it might also provide critical insights for designing targeted rehabilitation interventions. Importantly, to fully capture the dynamics of dual-task interference, both the motor and cognitive components will be evaluated under single- and dual-task conditions, ensuring a comprehensive assessment of reciprocal dual-task effects.

To the best of the authors' knowledge, this is the first study to systematically evaluate both short- and long-term effects of deep brain stimulation on motor-cognitive performance during walking using a dual-task paradigm and muscle synergy theory. This study aims to investigate whether treatment with DBS leads to improvements in dual-task capacity or results in different attentional strategies during walking with a concurrent cognitive task, thereby addressing a critical gap in the literature and offering novel insights into the interplay between cognitive and motor functions in patients with PD.

II. Material and Methods

A. Sample Population and Experimental Protocol

A total of 27 individuals with advanced PD (19 men, 8 women; age: 57.4 $\pm$ 1.5 years; height: 1.72 $\pm$ 0.02 m; body mass: 74.4 $\pm$ 2.7 kg) and 30 healthy controls with comparable age (12 men, 18 women; age: 55.0 $\pm$ 1.6 years; height: 1.68 $\pm$ 0.01 m; body mass: 74.1 $\pm$ 3.4 kg) participated in this study. The sample size was determined based on prior studies targeting similar endpoints in PD patients [14]. The PD cohort was recruited from the Stereotactic and Functional Neurosurgery Unit at the University of Turin (Italy) among candidates for bilateral, high-frequency (130 Hz) subthalamic nucleus DBS. PD patients were carefully selected using stringent inclusion criteria as described in [15]. Briefly, PD patients were included if they met UK Brain Bank criteria, had good levodopa response, no major cognitive, psychiatric, or gait-impairing comorbidities, were under 70 at DBS neurosurgery, and could walk unaided. Control participants were recruited among patients' caregivers (most frequently their partners, sharing a similar age), and excluded in presence of cognitive deficits and musculoskeletal disorders affecting gait. Both PD patients and healthy subjects underwent neurophysiological and behavioral assessment as detailed in [15]. The complete list of tests performed is provided in the **Table S1**.

PD patients were evaluated at three different time points: prior to DBS surgery (baseline, T_0) during their optimal "best-ON" pharmacological condition, and 3 months (T_1) and 12 months post-surgery (T_2) during their optimal "best-ON" pharmacological and stimulation conditions. By contrast, controls underwent a single experimental session.

Experimental sessions involved the execution of three distinct tasks:

- **Cognitive Single Task (cST)**: while seated, participants were instructed to perform a phonemic fluency task, consisting in generating as many words as possible beginning with the letter 'T' within a fixed 5-minute interval [18], [19]. The target letter was provided immediately before the task onset. All responses were verbalized in the participants' native language (Italian);
- **Motor Single Task (mST)**: participants were required to walk barefoot on a flat surface for 5 minutes at a self-selected pace [15]. The path consisted of a 9-meter straight section, which participants traversed back and forth, approaching the turns alternately in opposite directions;
- **Motor-cognitive Dual Task (mcDT)**: participants were required to walk as described in the mST condition while performing the same phonemic fluency task executed in cST. During mcDT, no explicit instructions regarding prioritization were given to participants. To minimize learning effects, target letters were alternated across dual-task sessions ('T' at T_0, 'R' at T_1, and 'T' at T_2).

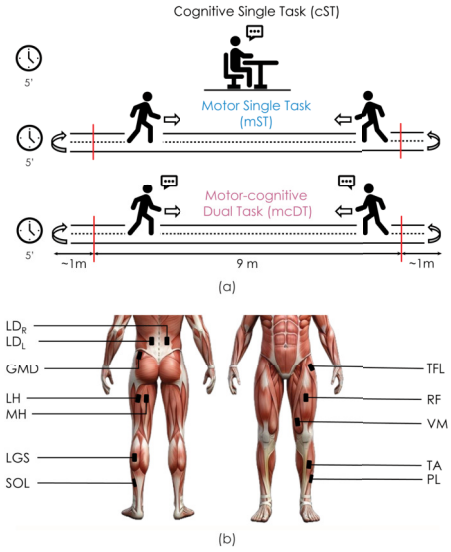

Fig. 1. (a) Schematic representation of cognitive Single Task (cST), motor Single Task (mST), and motor-cognitive Dual Task (mcDT). (b) EMG probe positioning on the following muscles: right (LD_R) and left (LD_L) Longissimus Dorsii, Tensor Fasciae Latae (TFL), Gluteus Medius (GMD), Rectus Femoris (RF), Lateral Hamstrings (LH), Medial Hamstrings (MH), Vastus Medialis (VM), Lateral Gastrocnemius (LGS), Peroneus Longus (PL), Soleus (SOL), and Tibialis Anterior (TA).

Additionally, a minimum interval of 6 months was maintained between the cST and mcDT sessions.

Fig. 1a schematically shows the three distinct tasks and the walking path.

The cST was performed first, on a separate day prior to the motor tasks. The two motor tasks (mST and mcDT) were conducted on the same day in a randomized order to mitigate potential fatigue-related effects in the assessment of motor performance. For the PD group, cST was conducted only at T_0 and T_2, while mST and mcDT were conducted at every time point (T_0, T_1, and T_2). To control for both patient- and task-related influences on dual-task interference, the exact same dual-task procedure was used across the three PD experimental sessions. The same set of instructions was provided each time, environmental distractions were minimized, and task difficulty was kept consistent.

The Motion Analysis Laboratory of the Polito^{BIO}Med Lab (Politecnico di Torino, Turin, Italy) hosted all the experimental sessions. The study protocol received approval from the Ethics Committee of A.O.U. Città della Salute e della Scienza di Torino – A.O. Ordine Mauriziano – A.S.L. "Città di Torino" (approval number 0092029, issued on September 11, 2018). In accordance with the Declaration of Helsinki, written informed consent was obtained from all participants prior to their involvement in the research.

B. Data Acquisition and EMG Pre-Processing

Surface electromyographic (EMG) signals were collected from 12 trunk and lower-limb muscles using the STEP32 system (Medical Technology, Turin, Italy), already employed in previous clinical gait analysis studies [20], [21]. Examined muscles included right and left Longissimus Dorsi (LD_R and LD_L), Tensor Fasciae Latae (TFL), Gluteus Medius (GMD), Rectus Femoris (RF), Lateral and Medial Hamstrings (LH and MH), Vastus Medialis (VM), Lateral Gastrocnemius (LGS), Peroneus Longus (PL), Soleus (SOL), and Tibialis Anterior (TA). **Fig. 1b** shows an example of EMG probe positioning. In PD participants, EMG signals were acquired from the more affected side (15 left, 12 right), while in healthy volunteers from the dominant side (27 right, 3 left). Additionally, gait cycle timings were detected by means of 3 foot-switch sensors placed under the heel, the first, and the fifth metatarsal head of each foot as detailed in [22]. Gait data were simultaneously recorded at a sampling rate of 2 kHz and then processed in MATLAB (version 2024a, MathWorks Inc., Natick, USA) using custom routines. Before muscle synergy extraction, EMG signals were pre-processed according to as described in a previous study [15]. Briefly, gait cycles were segmented using the foot-switch signals as described in a previous study [22]. Only gait cycles belonging to the straight path were included, discarding those walked during direction changes at the beginning and at the end of the straight walkway [23]. For the selected cycles, EMG envelopes were computed by applying an 8^{th}-order high-pass filter at 35 Hz, full-wave rectification, and a 4^{th}-order low-pass filter at 12 Hz. Each envelope was amplitude-normalized to the global maximum for each muscle. Finally, EMG envelopes were time-normalized to 1000 points (600 stance, 400 swing) to remove variability due to cycle duration. To assess changes in neuromuscular control over motor task duration, the amplitude- and time-normalized EMG envelopes were concatenated into 10-gait-cycle epochs prior to muscle synergy extraction.

C. Muscle Synergy Extraction

For each subject, motor task, and time point, the Non-Negative Matrix Factorization (NNMF) algorithm was applied to each 10-gait-cycle epoch to extract time-invariant muscle synergies. The NNMF algorithm models each 10-gait-cycle matrix as a linear combination of activation coefficients (C) and weight vectors (W). The latter models the time-independent contribution of each muscle to a specific muscle synergy, while the former describes the time-dependent modulation of each muscle synergy. More specifically, the following NNMF parameters were set: a maximum of 1000 iterations, a minimal residual value of $1e^{-6}$, and a tolerance of $1e^{-6}$ [24]. Instead of using the multiplicative update rule of the original NNMF algorithm [25], we chose the Non-Negative Least Squares (NNLS) implementation [26], due to its advantages in terms of reduced computational time.

The factorization algorithm was iteratively applied to each 10-gait-cycle epoch, changing the number of muscle synergies in the range [1], [12] in order to explore different solutions. The final number of muscle synergies (N) was then selected using the "elbow" criterion [27], based on identifying the point of maximum curvature in the R^2 vs. number of muscle synergies curve. For each subject, the representative number of synergies was defined as the mode of the values obtained across all 10-gait-cycle epochs belonging to the same participant. Subsequently, the number of synergies characterizing

each population, motor task (mST and mcDT), and time point (T_0, T_1, and T_2 for PD patients; single session for controls) was determined as the mode of all the subject-specific numbers of synergies extracted from the corresponding population, motor task, and time point. To obtain an average representation of the motor control strategies of each individual, as well as of each population, in each motor condition, and time point, the previously computed weight vectors were sorted using a *k*-means clustering algorithm set as detailed in [15]. The activation coefficients were then sorted accordingly.

D. Motor and Cognitive Performance

To evaluate the impact of simultaneous task execution on motor and cognitive performance, standard and muscle synergy-derived motor parameters and a cognitive parameter obtained from the phonemic verbal fluency task were analyzed.

1) Motor Performance: Gait speed and two synergy derived metrics, one from the activation coefficients (C) and one from the weight vectors (W), were computed

a) *Gait speed (Speed)*: defined as the total distance walked along the straight path (9 m) divided by the total time required to go through it [2];

b) *Full Width at Half Maximum (FWHM):* it quantifies the duration (or width) of a muscle synergy's activation profile, describing the period during which a group of muscles is actively engaged in a given task. After subtracting the minimum value from each data point throughout the gait cycle, FWHM is estimated as the cumulative duration (expressed as percentage of the gait cycle) during which the activation coefficient (C) exceeds half of its maximum [28]. FWHM is computed for each muscle synergy, separately, and then it is averaged across synergies to obtain a representative value for each subject's task. FWHM ranges from 0 to 100, with lower values indicating a short activation duration of the muscle synergy and higher values indicating a long activation duration;

c) *Weight Selectivity Index (WSI):* is a measure of motor control selectivity. More specifically, for each muscle, WSI_m quantifies the extent to which a single muscle is recruited by multiple muscle synergies, and it is computed by adapting the metric originally proposed by Hoyer [29] as follows:

$$WSI_m = \frac{\sqrt{N} - \frac{\|W_{m,syn}\|_1}{\|W_{m,syn}\|_2}}{\sqrt{N} - 1} \tag{1}$$

where $\|\cdot\|_1$ and $\|\cdot\|_2$ are the L_1-norm and L_2-norm operators, $W_{m,syn}$ is the m^{th}-muscle weight vector (with $syn = 1, N$), and N is the number of synergies. Then, WSI_m values are aggregated into four distinct clusters, separately averaging the muscles involved in load acceptance (VM, TFL, GMD, and RF), propulsion (LGS, PL, and SOL), trunk stabilization (LD_R and LD_L), and swing (MH, LH, and TA), respectively [24]. WSI ranges from 0 to 1, where 0 corresponds to a cluster of muscles equally distributed across N synergies and 1 represents the condition of maximum sparseness (i.e., when the cluster of muscles contributes exclusively to a single muscle synergy). Additionally, an overall average across clusters is calculated.

2) Cognitive Performance: The number of words (N_{words}) produced during the phonemic fluency task was counted, both in cST and mcDT conditions, by the same neuropsychologist [30].

E. Dual-Task Effect (DTE)

Following Plummer and Eskes [11], Dual Task Effect (DTE) is defined as:

$$DTE = \frac{Dual\ Task\ X - Single\ Task\ X}{Single\ Task\ X} \cdot 100 \tag{2}$$

where X represents the generic variable monitored in single- and dual-task conditions. Dual-task interference is then assessed by estimating at least a DTE for each of the two tasks examined (motor and cognitive). Considering the motor (X = Speed, X = FWHM, and X = WSI) and cognitive ($X = N_{words}$) variables defined in the previous paragraph, we calculated three DTEs for the motor task (named $mDTE_{Speed}$, $mDTE_{FWHM}$, and $mDTE_{WSI}$) and one DTE for the cognitive task (named cDTE) to quantify dual-task interference. By convention, positive DTE values indicate a relative improvement in performance during the dual-task conditions (i.e., dual-task benefit), while negative values indicate a decline in performance compared to the single-task conditions (i.e., dual-task cost) [11]. Accordingly, a negative sign was introduced for those variables initially not complying with this convention (e.g., FWHM) to ensure consistency:

$$DTE = \frac{-(Dual\ Task\ X - Single\ Task\ X)}{Single\ Task\ X} \cdot 100 \tag{3}$$

To integrate both motor and cognitive DTE into a single metric and better quantify the overall dual-task interference, the combined Dual-Task Effect (combDTE) was computed as defined by Longhurst et al. [31]. We computed $combDTE_{Speed}$, $combDTE_{FWHM}$, and $combDTE_{WSI}$; $combDTE_{FWHM}$ and $combDTE_{WSI}$ used FWHM and WSI averaged across synergies.

F. Statistical Analysis

Levene's F test was used to assess the homogeneity of variances for all data distributions [32]. Differences between groups (PD patients at different time points - T_0,T_1, and T_2 - and controls) were tested for each condition (single- and dual-task) and metric (motor and cognitive metrics, including DTEs) through a Mixed Model based on repeated measure one-way ANOVA for PD patients and a one-way ANOVA for controls. *Post-hoc* analyses with Bonferroni adjustments for multiple comparisons were then performed. Differences between single- vs. dual-task conditions were tested using paired *t*-tests. Moreover, one-dimensional Statistical Parametric Mapping (SPM1D) [33] was employed to test differences between activation coefficient profiles. Effect sizes related to significant differences were calculated as Cohen's d [34]. The significance level α was set at 0.05 and all variables were

TABLE I
DEMOGRAPHIC AND ANTHROPOMETRIC CHARACTERISTICS OF PD PATIENTS AND CONTROLS

		Gender	Age (years)	Weight (kg)	Height (m)	UPDRS-III (Best-ON condition)	Disease duration (years)	LEDD (mg)	Educational level
PDs	Before DBS (T_0)	8 F, 19 M	57.4 ± 7.8	74.4 ± 14.0	1.72 ± 0.10	19.4 ± 9.4***	11.2 ± 3.1	1354.5 ± 79.9***	12.8 ± 4.2
	12 months after DBS (T_2)		58.6 ± 7.8	N/A	N/A	10.2 ± 5.2***	12.3 ± 3.1	669.4 ± 65.0***	12.8 ± 4.2
Controls		18 F, 12 M	55.0 ± 8.8	74.1 ± 18.6	1.68 ± 0.05	N/A	N/A	N/A	13.3 ± 3.8

*Values are reported as mean ± standard deviation over the sample populations. Abbreviations: PD Parkinson's Disease; DBS Deep Brain Stimulation; M Males; F Females; UPDRS-III Unified Parkinson's Disease Rating Motor Subscale; LEDD Levodopa Equivalent Daily Dose; N/A Not Applicable. Statistically significant differences between groups are represented through asterisks (*** $p < 0.001$).*

expressed as mean ± standard deviation across the sample population. SPSS Statistical Software, version 28.0 (SPSS Inc., Chicago, IL), was used to perform all the statistical analyses.

III. RESULTS

The demographic and anthropometric characteristics of the PD (before DBS and 12 months after DBS) and controls enrolled in the study are detailed in **Table I**. No statistically significant differences were detected between PDs and controls for age, weight, height, and educational level. On average, PD patients showed clinical improvement in motor performance ($p < 0.0001$; $d = 1.22$) and reduced levodopa intake ($p < 0.0001$; $d = 9.41$) after DBS surgery.

A. Muscle Synergy Composition

For both sample populations included in the study (PD at the three time points - T_0, T_1, and T_2 - and controls), four muscle synergies were selected, on average, to properly model the original EMG data during both motor single task (PD at T_0: 4.2 ± 0.4 (mean ± standard deviation); PD at T_1: 4.2 ± 0.4; PD at T_2: 4.4 ± 0.7; Controls: 4.4 ± 0.6) and motor-cognitive dual-task (PD at T_0: 4.4 ± 0.6; PD at T_1: 4.1 ± 0.3; PD at T_2: 4.4 ± 0.7; Controls: 4.4 ± 0.7) conditions. No statistically significant differences in the number of muscle synergies were observed between groups or conditions.

Fig. 2 shows the composition of the average muscle synergies extracted during motor single task and motor-cognitive dual-task conditions for PD patients before DBS, at 3 months and 12 months after DBS surgery, and controls. Considering control subjects (**Fig. 2d**), four different biomechanical functions can be identified, one per muscle synergy [23]:

1) *Load acceptance*: modeled by the first muscle synergy and mainly characterized by the activation of the muscles contributing to movement and stabilization of the coxo-femoral joint (i.e., TFL and GMD), as well as quadriceps muscles (i.e., VM and RF) supporting knee extension;
2) *Propulsion*: represented by the second muscle synergy and mainly characterized by the activation of the LGS, PL, and SOL muscles, which contribute to foot plantarflexion and shank flexion to generate propulsion;
3) *Trunk stabilization*: modeled by the third muscle synergy and mainly characterized by the activation of the trunk muscles (i.e., the two LDs) contributing to trunk stabilization in terminal stance;
4) *Swing*: represented by the fourth muscle synergy and mainly characterized by the activation of TA, controlling ankle dorsiflexion to ensure proper foot clearance from the ground, and MH/LH muscles to decelerate the lower limb and prepare for the upcoming heel strike during the late-swing phase.

PD patients, at all time points (T_0, T_1, and T_2), demonstrated similar biomechanical functions. Although the number and composition of muscle synergies remain unchanged with the introduction of a second task across all sample populations, reorganization within the synergies is observed, mainly at the level of activation coefficients, as indicated by the transparent gray rectangles reported in **Fig. 2**.

B. Motor and Cognitive Performance

The detailed results about cognitive single-task, motor single-task and motor-cognitive dual-task parameters can be found in the Supplementary Material (**Table S2**).

1) Motor Performance:

a) *Gait speed*: No significant differences were found between PD patients (at any time point) and controls during either mST ($p = 0.31$) or mcDT ($p = 0.08$);
b) *FWHM*: **Fig. 3** represents the FWHM boxplots for each of the tested groups during motor Single Task (mST – **Fig. 3a**) and motor-cognitive Dual Task (mcDT – **Fig. 3b**). During mST condition, PD patients revealed significantly higher FWHM values ($p<0.001$) at all three time points compared to controls for all muscle synergies (and their total average), except for the one associated

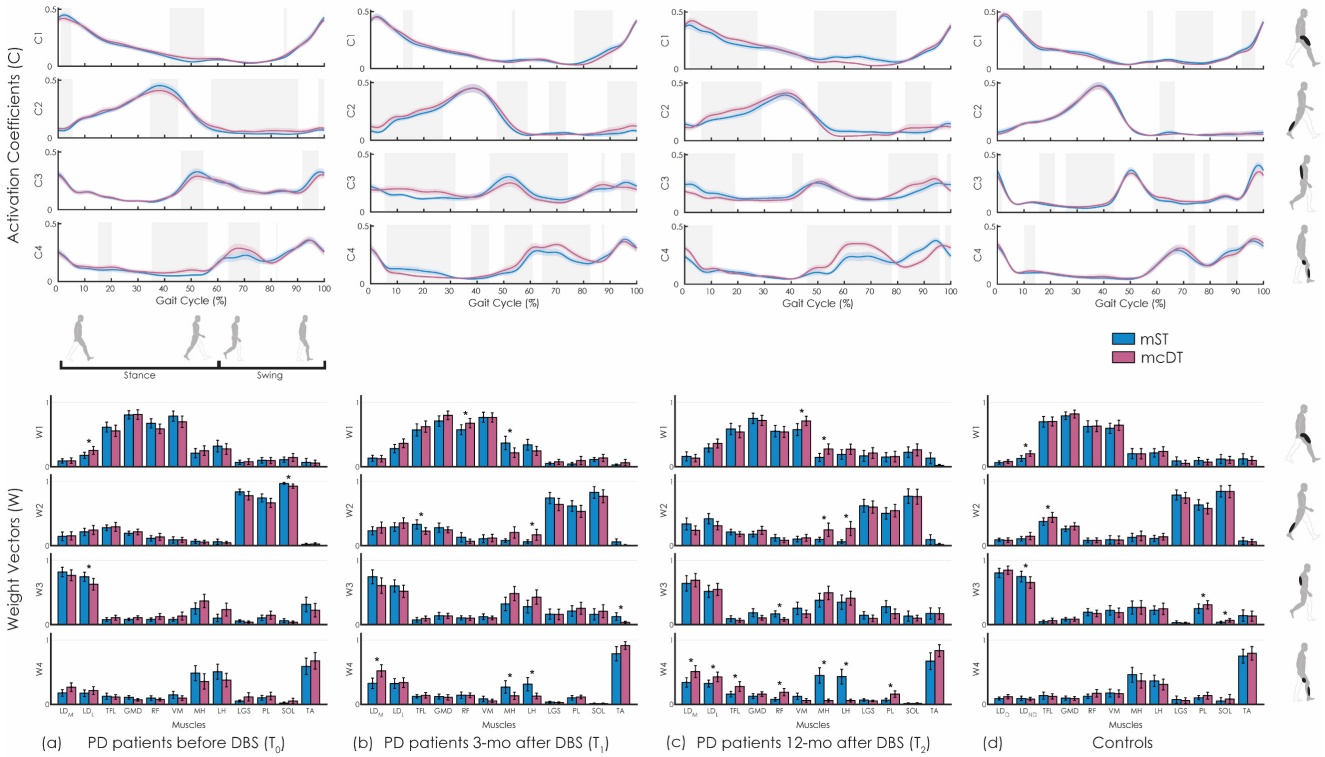

Fig. 2. Comparison of average muscle synergies between motor single task (blue) and motor-cognitive dual-task (pink) conditions for (a) PD patients before DBS, (b) at 3 months after DBS, (c) at 12 months after DBS, and (d) controls. At the top, the activation coefficients are shown as average curves with standard error bands across the sample population; gray transparent rectangles indicate significant differences between single- and dual-task conditions. At the bottom, the weight vectors are displayed with average values and standard errors; asterisks mark statistically significant differences between single- and dual-task conditions. Abbreviations: PD Parkinson's Disease; mST motor single task, mcDT motor-cognitive dual task, VM vastus medialis, TFL tensor fasciae latae, GMD gluteus medius, MH medial hamstring, LD longissimus dorsii (for PD patients: $LD_M \equiv$ LD of the more affected side, $LD_L \equiv$ LD of the less affected side; for controls: $LD_D \equiv$ LD of the dominant side, $LD_{ND} \equiv$ LD of the non-dominant side), TA tibialis anterior, LGS lateral gastrocnemius, PL peroneus longus, SOL soleus, RF rectus femoris, and LH lateral hamstring.

with Swing. Similar differences were observed during the mcDT condition ($p < 0.001$), with an additional statistically significant difference between PD patients at T_2 and controls for the Swing muscle synergy;

c) *WSI*: **Fig. 4** represents the WSI boxplots for each of the tested groups during motor Single Task (mST – **Fig. 4a**) and motor-cognitive Dual Task (mcDT – **Fig. 4b**). During motor single task and motor-cognitive dual task conditions, PD patients after DBS surgery revealed statistically lower WSI values compared to PD patients before DBS (mST: $p < 0.04$; mcDT: $p < 0.03$) and controls (mST: $p < 0.05$; mcDT: $p < 0.04$). This difference was observed for the muscle synergy involving trunk stabilization muscles, as well as for the average across all synergies.

2) Cognitive Performance: No significant differences were observed between PD patients (both at T_0 and T_2) and controls in terms of words produced during the phonemic verbal fluency task during either cognitive Single Task ($F(2,81) = 0.97$; $p = 0.39$) or Dual Task conditions ($F(2,81) = 1.25$; $p = 0.29$). However, all groups produced significantly fewer words when performing motor-cognitive dual task condition compared to cognitive single task condition (PDs at T_0: $t(26) = 3.25$, $p = 0.003$, $d = 0.62$; PDs at T_2: $t(26) = 2.85$, $p = 0.009$, $d = 0.55$; Controls: $t(29) = 5.51$, $p<0.001$, $d = 1.01$), showing a detrimental impact of the motor task on the cognitive performance.

C. Dual-Task Effect (DTE)

Analysis of $mDTE_{Speed}$ indicated that attention sharing reduced gait speed across all groups, as reflected by negative $mDTE_{Speed}$ values, with no statistically significant differences between groups (see **Table II**).

Fig. 3c and **Fig. 4c** show the distribution of the motor DTE computed from the FWHM ($mDTE_{FWHM}$) and WSI ($mDTE_{WSI}$) parameters, respectively. Analysis of $mDTE_{FWHM}$ indicates that the phonemic verbal fluency task has a detrimental effect on PD motor performance prior to DBS implantation, as shown by negative $mDTE_{FWHM}$ values (PD at T_0: $-12.5 \pm 11.5\%$). This effect is greater in PD patients before DBS compared to controls, in the propulsive muscle synergy and in the average across all synergies (see **Fig. 3c**). Following DBS surgery, the difference between PD patients and controls is no longer statistically significant. Regarding $mDTE_{WSI}$, no significant differences were observed between sample populations (see **Fig. 4c**); on average, the attention sharing negatively impacted motor performance across all subject groups. Average $mDTE_{FWHM}$, and $mDTE_{WSI}$ values are reported in **Table II** with the indication of the statistically significant differences among groups.

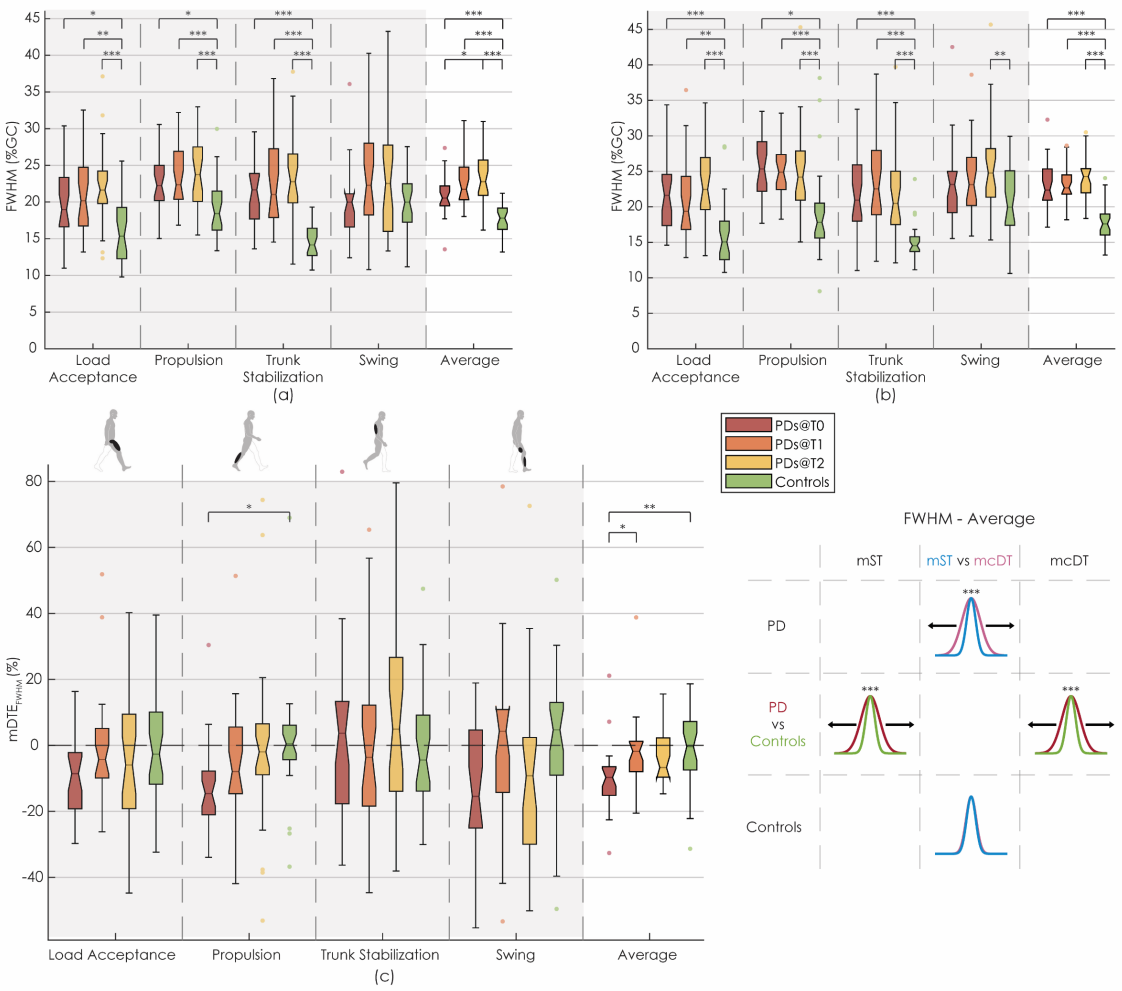

Fig. 3. Boxplots representing the Full Width at Half-Maximum (FWHM) for Parkinson's Disease (PD) patients before DBS (T_0, in red), at 3 months after DBS (T_1, in orange), at 12 months after DBS (T_2, in yellow), and controls (in green), during (a) motor single task (mST) and (b) motor-cognitive dual task (mcDT). Panel (c) shows the FWHM motor Dual Task Effect ($mDTE_{FWHM}$). Data are presented both separately for each of the four biomechanical functions (muscle synergies) and as averages across all muscle synergies. Abbreviations: %GC percentage of the Gait Cycle. Statistically significant differences are highlighted by asterisks ($^{*}p < 0.05$, $^{**}p < 0.01$, $^{***}p < 0.001$). The bottom right corner of the figure shows a graphical summary of the FWHM results averaged across muscle synergies.

Reciprocal DTEs were assessed by plotting cognitive and motor dual-task effects (DTE) against each other, as illustrated in **Fig. 5**. DTE domain can be divided into four distinct regions as it follows:

1. *Mutual interference*: both cognitive and motor performance deteriorate (negative DTEs in both tasks), indicating that attentional demands exceed available resources and neither motor nor cognitive task can be maintained at single-task performance levels;
2. *Gait prioritization*: only the cognitive task shows interference while motor performance is maintained, indicating that attentional resources are insufficient for both tasks, but the participant prioritizes the motor task, resulting in a decline in cognitive performance only;
3. *Motor facilitation*: both cognitive and motor performance improve (positive DTEs in both tasks), suggesting that attentional resources are sufficient to enhance performance in both tasks when performed together;
4. *Cognitive prioritization*: only the motor task shows interference while cognitive performance is maintained, again reflecting limited attentional resources. In this case, the participant prioritizes the cognitive task, leading to a decline in motor performance only.

As shown in **Fig. 5**, PD patients prior to DBS surgery (bold red square) exhibited a mutual interference pattern, with declines observed in both motor and cognitive tasks, as indicated by negative $mDTE_{FWHM}$ and cDTE values, reflecting a "posture-second" strategy ($mDTE_{FWHM}$: $-12.5 \pm 11.5\%$; cDTE: $-12.4 \pm 23.2\%$). At 12 months post-DBS (bold yellow triangle), these patients continued to display a mutual interference pattern; however, they showed a substantial improvement in motor dual-task effect compared to baseline ($mDTE_{FWHM}$: $-4.5 \pm 8.3\%$), while cognitive performance remained stable (cDTE: $-11.2 \pm 25.9\%$). This suggests an overall reduction in dual-task interference after surgery. In contrast, control participants (bold green circle) primarily prioritized gait performance

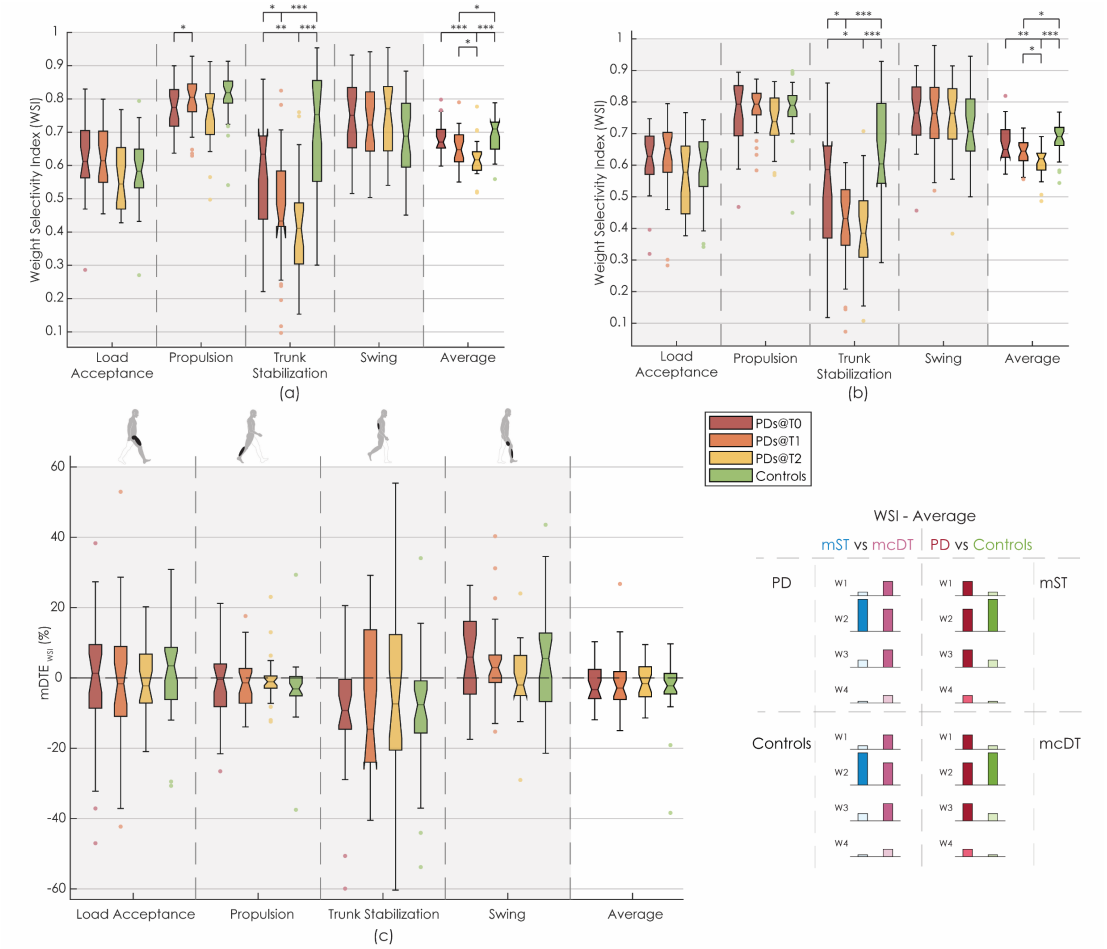

Fig. 4. Boxplots representing the Weight Selectivity Index (WSI) for Parkinson's Disease (PD) patients before DBS (T_0, in red), at 3 months after DBS (T_1, in orange), at 12 months after DBS (T_2, in yellow), and controls (in green), during (a) motor single task (mST) and (b) motor-cognitive dual task (mcDT). Panel (c) shows the WSI motor Dual Task Effect ($mDTE_{WSI}$). Data are presented both separately for each of the four biomechanical functions (muscle synergies) and as averages across all muscle synergies. Statistically significant differences are highlighted by asterisks (*p<0.05, **p<0.01, ***p<0.001). The bottom right corner of the figure shows a graphical summary of the WSI results averaged across muscle clusters.

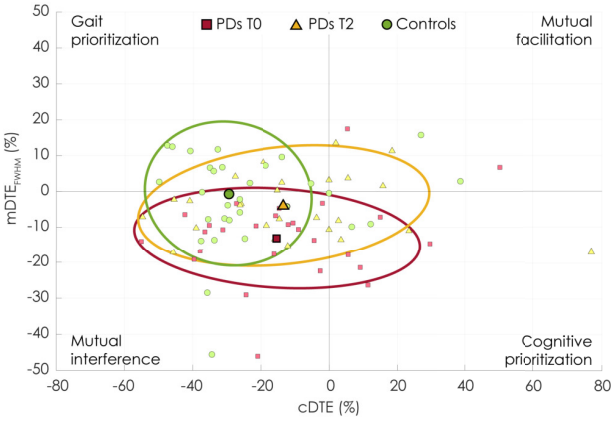

Fig. 5. Interaction between motor (mDTE) and cognitive (cDTE) dual-task effects for Parkinson's Disease (PD) patients before DBS (T_0, red), at 12 months after DBS (T_2, yellow), and controls (green). Circles, squares, and triangles represent the mDTE and cDTE values for each participant. Colored ellipses indicate the 95% confidence region for each group. Bold circle, square, and triangle represent groups' centroid.

($mDTE_{FWHM}$ = −1.1 ± 12.7%), often at the expense of cognitive performance (cDTE: −23.0 ± 21.0%), reflecting a "posture-first" strategy.

Because $mDTE_{WSI}$ did not show statistically significant differences among groups, only $mDTE_{FWHM}$ was used for assessing reciprocal DTEs.

No statistically significant between-group differences in combDTEs were found for Speed, FWHM, or WSI. Dual-task walking generally reduced motor automaticity in each group, as indicated by negative combDTEs (see **Table II**).

IV. Discussion

To the best of the authors' knowledge, this is the first study to evaluate both short- and long-term effects of high-frequency deep brain stimulation of the subthalamic nucleus on motor and cognitive performance using both dual-task paradigm and muscle synergy theory. This study aimed to investigate whether DBS treatment leads to improvements in dual-task capacity or results in different attentional strategies during walking while performing a concurrent cognitive task.

High-frequency bilateral subthalamic stimulation has already been associated with improvements in single-task gait performance, such as stride length, walking speed, and rhythmicity [9], [35]. However, little is known about the

TABLE II
MOTOR AND COGNITIVE DUAL-TASK EFFECT (DTE) FOR PD PATIENTS AND CONTROLS

Motor Dual-Task Effect (mDTE)		**PD Patients**			**Controls**
		Before DBS (T_0)	**3 Months after DBS (T_1)**	**12 Months after DBS (T_2)**	
$mDTE_{Speed}$ (%)		-16.6 ± 12.3	-14.0 ± 6.7	-12.9 ± 9.1	-11.3 ± 9.7
$mDTE_{FWHM}$ (%)	Load Acceptance	-12.7 ± 13.9	-3.2 ± 15.0	-9.1 ± 25.3	-2.4 ± 18.9
	Propulsion	-16.0 ± 13.3*	-8.3 ± 18.7	-6.9 ± 30.6	-0.3 ± 18.9*
	Trunk Stabilization	-4.5 ± 23.4	-5.4 ± 28.1	2.6 ± 25.5	-4.0 ± 16.4
	Swing	-22.1 ± 31.5	-6.5 ± 30.9	-18.0 ± 32.7	-4.0 ± 26.7
	Average	-12.5 ± 11.5†,‡	-3.7 ± 10.2†	-4.5 ± 8.3	-1.1 ± 12.7‡
$mDTE_{WSI}$ (%)	Load Acceptance	1.3 ± 18.8	-1.7 ± 18.8	-2.3 ± 16.6	3.4 ± 14.8
	Propulsion	-0.5 ± 10.7	-1.4 ± 7.1	-1.1 ± 6.9	-3.1 ± 9.2
	Trunk Stabilization	-9.3 ± 37.6	-14.6 ± 33.9	-7.4 ± 27.3	-7.6 ± 27.1
	Swing	5.9 ± 13.3	2.9 ± 12.1	-2.0 ± 9.7	5.5 ± 15.2
	Average	-3.3 ± 5.9	-2.9 ± 8.4	-1.6 ± 5.6	-2.3 ± 5.7
Cognitive Dual-Task Effect (cDTE)					
cDTE (%)		-12.4 ± 23.2	N/A	-11.2 ± 25.9	-23.0 ± 21.0
Combined Dual-Task Effect (combDTE)					
$combDTE_{Speed}$ (%)		-27.2 ± 20.9	N/A	-22.0 ± 23.1	-32.0 ± 19.0
$combDTE_{FWHM}$ (%)		-15.0 ± 23.7	N/A	-11.0 ± 25.9	-23.2 ± 21.4
$combDTE_{WSI}$ (%)		-13.5 ± 24.0	N/A	-11.44 ± 23.6	-24.3 ± 20.7

Values are reported as mean ± standard deviation over the sample populations. Statistically significant differences ($p<0.05$) between groups (PD patients at different time points - T_0, T_1, and T_2 - and controls) are represented by means of asterisks (*), daggers (†), and double daggers (‡). Abbreviations: N/A Not Assessed; DBS Deep Brain Stimulation; FWHM Full-Width at Half Maximum; $mDTE_{FWHM}$ FWHM-derived motor dual-task effect; $mDTE_{WSI}$ WSI-derived motor dual-task effect; $mDTE_{Speed}$ Speed-derived motor dual-task effect; cDTE cognitive dual-task effect; $combDTE_{Speed}$, $combDTE_{FWHM}$, and $combDTE_{WSI}$ are the combined dual-task effects derived from Speed, FWHM, and WSI parameters, respectively.

effects of this stimulation on dual-task walking, a crucial component of daily mobility. We hypothesized that, after DBS surgery, dual-task capacity in patients with Parkinson's disease would improve, reflecting a reduced cognitive load required to perform simultaneous motor and cognitive tasks. This expectation was based on previous literature suggesting that DBS would enhance gait automaticity and rhythmicity [35].

Results supported our original hypothesis, revealing an improvement in overall dual-task capacity following DBS surgery. Specifically, 12 months after surgery, PD patients showed a diminished impact of dual-tasking on motor performance without any detrimental effects on cognitive performance, becoming no different from that of healthy controls.

PD patients and controls were carefully matched on anthropometric and neuropsychological characteristics (e.g., height, weight, educational levels, etc.) to minimize confounding variables, thereby facilitating a more precise evaluation of dual-task interference effects.

Comprehensive locomotor control analysis through muscle synergy theory revealed both common patterns and differences between motor single-task (mST) and motor-cognitive dual-task (mcDT) conditions, as well as between PD and control subjects. Muscle synergy analysis revealed preserved structural organization (i.e., number and composition) across both groups and task conditions, despite significant differences in muscle synergy modulation (i.e., activation timing and muscle recruitment patterns) between mST and mcDT conditions. Concerning between-task differences in the activation coefficients, PD subjects exhibited longer muscle activation durations (i.e., greater FWHM values) across all synergies compared to control subjects, suggesting the need for a more robust locomotor control to cope with the increased motor control system's degrees of freedom due to the mcDT condition [36], [37]. This finding aligns with the work by Janshen et al. [38], who reported larger FWHM values in patients suffering from multiple sclerosis compared to healthy controls as a compensatory strategy enhancing motor control robustness against pathology-related deficits. Similarly, D'Emanuele and colleagues [39] demonstrated a widening of time-dependent activation patterns in older adults during unstable, treadmill walking compared to stable walking to cope with external perturbations. Concerning between-task differences in the weight vectors, PD patients revealed significant modifications of trunk muscle activation strategies, characterized by reduced activation selectivity (i.e., lower WSI values) of the muscles belonging to the same muscle synergy. This alteration in motor control may be attributed to increased trunk muscle rigidity, a distinctive characteristic of PD pathophysiology [40], [41]. Specifically, PD patients after DBS surgery (at T_1 and even more at T_2) likely prioritized a reliable frontal-plane trunk stabilization through consistent

recruitment of an increased number of muscles over being more energetically efficient [42].

From a neurophysiological perspective, locomotion is governed by two distinct cortical networks: (i) a direct motor pathway (composed by the primary motor cortex, the cerebellum, and the spinal cord) managing automatic walking control, and (ii) an indirect motor pathway (composed by the prefrontal cortex and the premotor area) overseeing executive processes [43]. The preserved muscle synergy number (and composition) across conditions demonstrated in the present study may then suggest intact automatic control mechanisms, while the observed temporal and spatial synergy modulations likely reflect adaptations in the indirect locomotor pathway that modify these underlying motor commands [44].

Comparative analysis of muscle synergy-derived parameters revealed more pronounced differences between PD patients (at all timepoints) and controls in the activation coefficient metric (i.e., FWHM) than the weight vector metric (i.e., WSI). These findings align with the proposed neuroanatomical substrates of the muscle synergy components [45]. On the one hand, activation coefficients, which demonstrated significant alterations in PD patients, are hypothesised to originate from the supraspinal circuits, including motor cortex and basal ganglia networks directly compromised by dopaminergic degeneration characteristic of PD [46]. On the other hand, weight vectors, which exhibited minimal alterations in PD patients, are hypothesised to reflect spinal motor organization (not compromised by PD). These results support the concept that PD primarily affects higher-order motor control circuits while largely sparing spinal motor control mechanisms.

Results on dual-task effects revealed that PD patients at baseline employed a mutual interference strategy (or "posture-second" approach) during increased cognitive load conditions, allocating attention equally to both tasks and consequently experiencing significant decrements in both motor and cognitive performance due to limited attentional resources. This approach potentially compromises balance maintenance and increases fall risk in daily activities [7], [47]. DBS surgery substantially mitigated the phonemic task's impact on motor performance, as evidenced by the average FWHM measures at three months post-surgery ($mDTE_{FWHM} = -3.7 \pm 10.2\%$), with benefits persisting at 12 months ($mDTE_{FWHM} = -4.5 \pm 8.3\%$). Post-DBS patients demonstrated reduced dual-task costs across both cognitive and motor domains, confirming DBS's beneficial effect on dual-task capacity. In contrast, healthy controls primarily focus on locomotor performance at the expense of the cognitive task (reduced word production). This gait prioritization (or "posture-first") strategy is likely adopted by healthy adults due to heightened fear of falling, thereby protecting themselves from falls that would result in significant physical consequences [48], [49]. As previous studies have shown that dynamic balance and fall risk in PD patients are related to dual-task performance [50], [51], [52], the improvement in dual-task capacity observed in this study may also correspond to a reduced risk of falls and an improved dynamic balance after DBS surgery. This association could be explained by more effective management of attentional resources and improved ability to cope with challenging conditions following the intervention.

Notice that our interpretations based on muscle synergy–derived parameter (i.e., $mDTE_{FWHM}$) were supported by a conventional spatiotemporal gait measure (i.e., $mDTE_{Speed}$), which likewise suggest a reduced motor dual-task cost in PD patients after DBS surgery, although gait-speed differences did not reach statistical significance.

Unexpectedly, although not reaching statistical significance, PD patients showed reduced cognitive dual-task effects (cDTE) compared to controls. Nevertheless, both groups displayed comparable neuropsychological test performance and educational levels (**Table I**). These slight differences in dual-task behavior may be attributed to PD patients' greater familiarity with neuropsychological testing procedures routinely administered during clinical evaluations, potentially conferring an adaptive advantage during the experimental dual-task paradigm.

Results presented in this study differ from those reported by Seri-Fainshtat et al. [10], who found that dual-task costs for various aspects of gait did not improve following subthalamic nucleus DBS, suggesting that the surgical intervention does not enhance the automaticity of walking. However, their analysis focused exclusively on the motor dual-task effect, without assessing the cognitive dual-task effect. As a result, a comprehensive evaluation of dual-task interference was not possible in their study, which may explain the discrepancies with our findings.

Notice that, in this study, all PD patients were assessed under their optimal pharmacological and/or stimulation conditions. This approach was selected to better reflect patients' everyday functional status and because, for many participants, independently walking for five minutes in the OFF medication state would have been very difficult or even impossible. Furthermore, previous studies have reported no statistically significant differences in motor dual-task effects between stimulation ON and OFF conditions, or between medication ON and OFF conditions [10].

Although phonemic verbal fluency is among the most commonly used tasks to assess cognitive performance, it may present some limitations, as individuals can slow their responses to maintain accuracy during dual-task conditions. This could make it more challenging to detect subtle changes in dual-task performance. Future studies might consider including response rate measurement as well as more than one dual-task combination to address this potential issue. A second limitation of this study is the fact that no single-task cognitive evaluation (cST) was performed in PD patients three months after DBS surgery. According to the experimental protocol approved by the local ethics committee, the cST condition was included in the clinical assessments conducted by the neuropsychologist and neurosurgeon. However, these clinical evaluations took place only at baseline (T_0) and one year after surgery (T_2), but not at the three-month follow-up (T_1). A further limitation is the gender distribution difference between PD and control cohorts, due to fact that the disease is more prevalent in men than women, and that we recruited

controls of similar age among patients' caregivers (mainly their partners). However, height and weight did not differ significantly between cohorts, reducing the risk of confounding effects. Lastly, focusing solely on FWHM may limit insight into dual-task interference; future work should add metrics of both motor control and behavioral performance to refine these interpretations.

V. Conclusion

This study provides a comprehensive assessment of both short- and long-term effects of subthalamic nucleus deep brain stimulation on dual-task performance in Parkinson's disease patients using muscle synergy analysis. Our findings demonstrate that DBS significantly enhances dual-task capacity, with patients transitioning from a detrimental "posture-second" strategy at baseline to a more efficient attentional allocation pattern post-surgery. The reduced impact of cognitive loading on motor function without compromising cognitive performance suggests enhanced attentional resource management that may translate to improved dynamic balance and reduced fall risk in daily activities.

Acknowledgment

The authors sincerely thank all the volunteers for their invaluable participation and contribution to this study. This article reflects only the authors' views and opinions, and the Ministry cannot be considered responsible for them.

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