Dual-tasking reveals the attentional cost of resolving sensory conflict induced by perturbed optic flow during treadmill walking

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Highlights:

- Cognitive performance was preserved under dual-task walking for all conditions
- ML optic flow perturbations increases in kinematic and neuromuscular variability
- These effects are reduced in dual task conditions
- Individuals shift attention from the walking task to the cognitive task
- Resolving conflicting visual cues is attentionally costly

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Abstract:

The precise role of cognitive control in the processing of optic flow has been rarely investigated. Therefore, this study aimed to determine whether coping with unreliable visual inputs during walking requires cognitive resources.

Twenty-four healthy young adults walked on an instrumented treadmill in a virtual environment under two optic flow conditions: normal (congruent) and perturbed (continuous mediolateral pseudo-random oscillations). Each condition was performed under single-task and dual-task conditions of increasing difficulty (1-, 2-, 3-back). Foot placement kinematics (200 Hz) and surface electromyography (EMG) of *soleus* and *gluteus medius* (1000 Hz) were recorded. Means, standard deviations (variability), statistical persistence and step-to-step error correction were computed from gait time series in lateral and anteroposterior directions. For EMG variables, duration and variability of muscle activation were calculated from the full width at half maximum (FWHM) and the variance ratio, respectively. N-back task performance was expressed by the d prime and response time.

Cognitive performance decreased as cognitive load increased ($p<.001$) but remained preserved under dual-task walking. Kinematics variability and EMG variance ratio increased under optic flow perturbation ($p<.001$). However, dual-tasking reduced the impact of the optic flow disturbance on the kinematics variability. Persistence of step width and antipersistence of step velocity decreased as cognitive load increased. Lastly, FWHM of *soleus* muscle increased with dual-task ($p=.01$). The results indicated that cognitive dual-tasking decreased the optic flow effect, demonstrating that resolving conflicting visual cues was attentionally costly. Also, in dual-task conditions, individuals adopted a rigid and controlled walking pattern in order to succeed N-back task.

Keys words: Gait; Lateral Balance; Muscle activity; Executive functions; Detrended fluctuations analysis (fractal analysis).

Abbreviations:

1-back task (1b)
2-back task (2b)
3-back task (3b)
Analysis of variance (ANOVA)
Central Nervous System (CNS)
d prime (d')
Detrended Fluctuations Analysis (DFA)
DFA scale exponent (α)
Dual Task Cognitive (DTC)
Dual Task Walking (DTW)
False Alarm (FA)
Full Width at Half Maximum (FWHM)
Gluteus Medius (Gmed)
Interdisciplinary Center for Virtual Reality (CIREVE)
Lateral body position (z_B)
Linear correlation (R^2)
Mediolateral (ML)
Multidimensional scale NASA Task Load Index (NASA-TLX)
Normal Optic Flow (NOF)
Optic Flow (OF)
Perturbed Optic Flow (POF)
Preferred Walking Speed (PWS)
Raw NASA-TLX (R-TLX)
Response Time (RT)
Single Task Cognitive (STC)
Single Task Walking (STW)
Slope of least-squares fits (M)
Soleus (Sol)
Standard Deviation (SD, σ)
Step velocity (V)
Step width (W)
Surface electromyography (EMG)
Variance Ratio (VR)

1. Introduction

Walking is usually considered to be a simple and automatic behaviour relying on the spinal rhythm generators, present in the spinal cord (Brown, 1911; Takakusaki, 2017). However, walking in complex environments of everyday life requires the brain to continuously adapt to both external sensory inputs such as obstacles or irregular surfaces (Kuhman et al., 2021.) and to internal cognitively demanding tasks (Malcolm et al., 2018). To this end, human locomotion relies on a widespread cerebral network with cortical sensorimotor areas controlling gait initiation (Takakusaki, 2013, 2017) and subcortical structures such as basal ganglia ensuring gait regulation (Grillner et al., 2008; Hamacher et al., 2015). These mechanisms allow the control of multiple joints and muscle groups to cope with the associated large degrees of freedom (Bernstein, 1967).

As a way to improve the understanding of how the brain controls gait under unstable conditions, mediolateral (ML) perturbations (of the treadmill or optic flow) are of particular interest. Human walking is indeed more variable and unstable in the ML direction (Bauby and Kuo, 2000), in particular the step width and ML foot placement (O'Connor and Kuo, 2009; Bruijn and van Dieën, 2018; Dingwell and Cusumano, 2019). Recently, Dingwell and Cusumano (2019) proposed a ML multi-objective control of normal human gait and showed that the system primarily controls step width and, in a lesser extent, the ML displacement of the centre of mass to regulate walking. These results were latter experimentally confirmed under visually or physically disturbed walking (Kazanski et al., 2020, 2021; Render et al., 2021). In this context, step width and ML displacement of centre of mass were more variable but more tightly regulated when walking with continuous lateral oscillations of the visual field, and even more when walking with continuous lateral oscillations of the support, in comparison with no perturbation (Kazanski et al., 2020). However, there were no differences between healthy young and older adults, demonstrating that healthy ageing did not alter step-to-step ML regulation during perturbed walking. Since age-related cognitive decline led to lower adaptability under dual-task walking in older people (Beauchet et al., 2003; Al-Yahya et al., 2011; Mirelman et al., 2012), the likely effect of cognitive load on step-by-step ML regulation during perturbed walking remains an open question.

Francis et al. (2015) reported small effects of attentional sharing on mean and variability of step width and step length compared to the effects of optic flow perturbations. Under dual-task conditions, the attentional cost associated with the integration of gait perturbation depends on many factors including the type and complexity of the two tasks at hand (Al-Yahya et al., 2011). For example, cognitive tasks involving both working memory and information processing or attention have been shown to deteriorate performance and have greater impact on gait (*e.g.* variability of step width and stride time) than those involving inhibition or visuospatial cognition, especially during perturbed walking (Kao and Pierro, 2022). For instance, tasks that involved interfering factors at the external level (*e.g.* Go/No go task, choice reaction time task) have less impact on gait than those that involved interfering factors at the internal level (*e.g.* mental arithmetic task; n-back task, PASAT). This result is due to the competition between the cognitive functions involved in both locomotor and cognitive tasks which interfered with the self-organizing dynamics of the motor system (Lindenberger et al., 2000; Lövdén et al., 2008; Verrel et al., 2009). Using an incremental cognitive load protocol with young adults, a decrease in the antipersistence of the fluctuations of step velocity have been demonstrated (Decker et al., 2013) while Verrel et al. (2009) observed an increase in the regularity of walking. In contrast, in older adults, Verrel et al. (2009) found an increase in the regularity of walking at a low difficulty level because an easy cognitive task allows the system to easily perform the motor task in an automated fashion (Wulf and Prinz, 2001; Beilock et al., 2004). However, they observed a decrease at higher difficulty levels due to a cross-domain resource competition. Hence, this inverse U-shaped relationship occurred when participants reached their attentional capacity limit (Huxhold et al., 2006; Van Snellenberg et al., 2014). Malcolm et al. (2018) also showed that their young participants exhibited a decrease in the variability of step width and standard deviation of the head position in both directions under dual-task conditions. The authors interpreted this finding as a decrease of the visual perturbations' effects with the addition of a concurrent cognitive task. Similarly, Pechtl et al. (2020) found an increase of errors in a reaction time task performed in the visual modality under dual-task conditions when participants were exposed to optic flow perturbations during treadmill walking, while no increase was observed in the auditory modality. This allowed to confirm that the sensory modality of the concurrent task matters. Indeed, a cognitive task presented in the auditory modality may reduce the ability to process visual information (Wickens, 2002; Redfern et al., 2017, Brockhoff et al., 2022), due to the allocation of attentional resources towards another sensory (auditory) modality. In turn, this phenomenon could decrease the influence of ML optic flow perturbations on gait stability (e.g., standard deviation of the head position, Malcolm et al., 2018).

All movement, including gait, are initiated by the activation of one or more muscle groups. At this level, Acuña et al. (2019) showed that antagonist leg muscle coactivation is more sensitive to the effects of external optic flow perturbations than attentional sharing. In general, dual-task walking reduces muscle activity (Fraser et al., 2007) and the cortical contribution to this activation (Clark et al., 2013). In some cases, it also prevents the development of adaptation strategies during perturbed walking under single-task condition (Wellinghoff et al., 2014). In addition, the activation and power generated by the hip abductor (*gluteus medius*) and ankle plantar flexors (*soleus* and *gastrocnemius*) strongly contribute to the control of frontal plane balance and of foot placement in both AP and ML directions during walking (Neptune and McGowan, 2016; Roelker et al., 2019). An increased *gluteus medius* activity is predictive of a more lateral foot placement (Rankin et al., 2014) and has been associated with an increased step width (Kubinski et al., 2015). These results have been confirmed by Stokes et al. (2017) during walking under optical flow perturbations. By using perturbations of the support surface during walking (*i.e.* respectively slippery floor or beam and uneven surface or platform displacements), Martino et al. (2015) and Santuz et al. (2020) have observed an increase of muscle activation duration of the lower limbs, which is interpreted as a compensatory mechanism implemented by the CNS under increased postural threat. To our knowledge, no study has assessed the extent to which the availability of attentional resources modulates this compensatory mechanism.

Based on the theoretical and computational framework of a ML multi-objective control model of human gait Dingwell and Cusumano (2019), the present study aimed to explore the attentional cost associated with gait control under ML optic flow perturbations at both the kinematic and muscular levels in young adults. We assumed that attentional sharing would lower the effects of ML optic flow perturbations on gait control due to the redirection of attentional resources towards the sensory (auditory) modality in which the concurrent task to walking is presented, and that this phenomenon would be more pronounced as the cognitive (working memory) load increases.

2. Materials and methods

2.1. Participants

Twenty-four healthy young adults (21.67 ± 2.28 years; 12 males and 12 females) took part in this study. Prior to the experiment, participants were interviewed regarding their health history. Non-inclusion criteria were: (1) any lower limb injury within the last six months, (2) pain, functional limitations or any discomfort that may affect walking, (3) history of vestibular, neurological or musculoskeletal disorders, (4) body mass index of 30 and higher, (5) sedentary lifestyle, (6) any medication altering the cognitive or physical abilities. The physical activity level was assessed by an adapted version of the question 1 of the Modifiable Activity Questionnaire (Vuillemin et al., 2000). The lower limb laterality was determined by the leg used to kick a ball (van Melick et al., 2017). All participants gave their written informed consent prior to the experiment. The study was approved by an Ethics Committee (IRB00012476-2020-30-11-76) and conducted with respect to the principles of the Declaration of Helsinki regarding human experimentation.

2.2. Experimental procedures

The experiment was conducted in the immersive room (dimensions: 4.80 × 9.60 × 3 m) of the Interdisciplinary Centre for Virtual Reality (CIREVE) at the University of Caen Normandie. The training session started with a familiarization of the participants to the 2 × 0.5 m walking treadmill (M-Gait, Motekforce Link, The Netherlands) followed by the determination of their preferred walking speed (PWS) according to the method of Jordan et al. (2007). Briefly, each participant started at a relatively low speed (*e.g.* 0.6 $m.s^{-1}$) which was progressively increased by 0.05 $m.s^{-1}$ increments until the participant reported being at his/her PWS. Then, walking speed was further increased by approximately 0.42 $m.s^{-1}$ and decreased in 0.05 $m.s^{-1}$ decrements until the participant reported once again being at his/her PWS. The procedure was repeated until the ascending and descending speeds reported by the participant were close (difference below 0.1 $m.s^{-1}$). Afterwards, the participants were familiarised with the auditory N-back task with three levels of working memory load (*i.e.* 1-back, 2-back, and 3-back). Each familiarization trial lasted at least 1 minute.

The testing session was composed of three blocks performed in a randomised order: (1) three N-back tasks in a seated position (single-task cognitive conditions, STC), (2) walking under unperturbed (normal) optic flow conditions (NOF), and (3) walking under perturbed (oscillating) optic flow conditions (POF) in the ML direction. In the latter two blocks, the walking tasks were performed under both single-task (STW) and dual-task (DTW) conditions (*i.e.* responding to the N-back tasks). To address the cognitive task, we used the cognitive dual-task (DTC). Participants were asked to walk naturally while looking straight ahead. The treadmill speed was adjusted to their PWS. The blocks (2) and (3) began and ended with a STW condition while the three DTW conditions were performed in a randomised order between the two STW conditions (Schaefer et al., 2015). Under dual-task conditions, no task priority instructions were given (Schaefer et al., 2015). Each condition lasted for 3 minutes. A total of thirteen experimental conditions were performed (Figure 1).

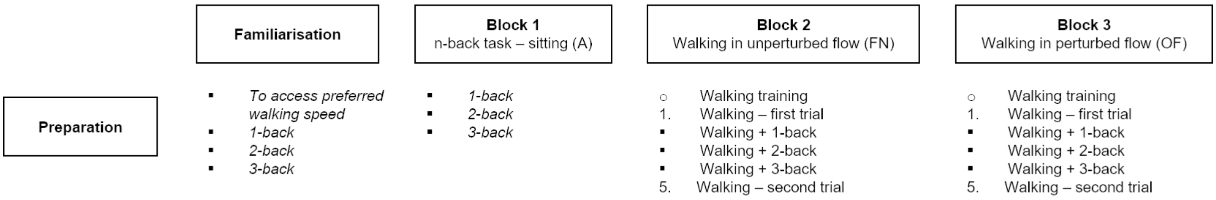

Figure 1. Experimental procedures including three experimental blocks. Empty bullets indicate training conditions while filled circles correspond to experimental conditions. Three blocks were performed in a randomised order. In the latter two blocks, participants began and ended with a locomotor single task condition (free walking) while the three dual task conditions were conducted in a randomised and counterbalanced order between both single tasks.

At the end of each condition, participants were asked to complete the NASA Task Load Index (NASA-TLX), a subjective multidimensional assessment questionnaire of perceived cognitive load, on a digital tablet (Hart and Staveland, 1988; French version: Cegarra and Morgado, 2009). In this questionnaire, three sub-scales relate to the demands imposed on the participant (*i.e.* physical, mental and temporal demands), and three others to the interaction of the participant with the task (*i.e.* effort, frustration, performance).

2.2.1. Manipulation of the optic flow

Participants were asked to walk on a street in a virtual environment of ancient Rome in which 2.4 m white statues were spaced every 3 meters to increase motion parallax (Bardy et al., 1996; Salinas et al., 2017; Figure 2). In all conditions, the antero-posterior velocity of the OF was congruent with the treadmill speed. During the perturbed OF conditions, the visual field oscillated in the ML direction. The perturbation consisted of a pseudo-random sum of four sinusoids with an amplitude of 0.25m that has previously been used in the literature (McAndrew et al., 2011; Kazanski et al., 2020):

$$D(t) = 0.25.[1.0 \sin(0.16.2\tau t) + 0.8 \sin(0.21.2\tau t) + 1.4 \sin(0.24.2\tau t) + 0.5 \sin(0.49.2\tau t)],$$

where D(t) is the lateral translation distance (m).

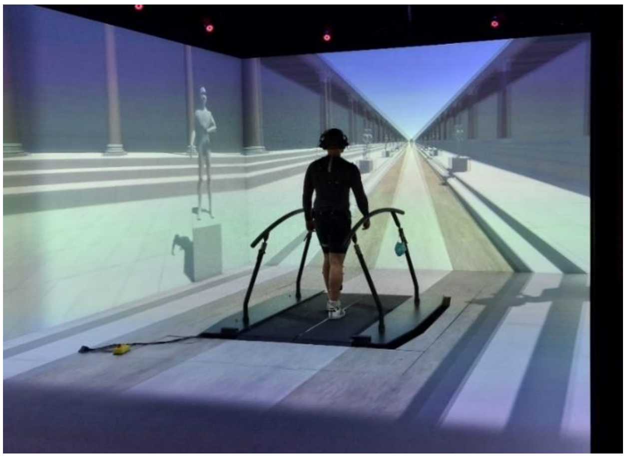

Figure 2. Experimental setup. Photograph of the V-Gait system (Motekforce Link, Amsterdam, Netherlands) used. The participant, equipped with EMG electrodes and retroreflective markers, is walking on the dual-belt instrumented treadmill in the virtual environment.

2.2.2. Manipulation of the cognitive load

The cognitive (working memory) load was parametrically manipulated through three levels of an auditory N-back task. N-back tasks involve the continuous sequences of auditory stimuli, presented one-by-one, in which the participant must determine if the currently presented stimulus is the same as the stimulus presented N trials before. Thus, the factor 'N' (number of items stored in working memory) allows increasing the task difficulty and thus the working memory load (Grissmann et al., 2017). By means of noise-cancelling headphones, participants heard easily and distinguishably the letters "E - A - I - O - R - T" pronounced in French and were asked to answer "Yes" when the letter heard was the same as the one presented N trials before (1-back, 2-back, 3-back). Responses were recorded using a microphone. The letters were presented with an inter-stimulus interval of 1800 to 2200 ms (2000 ± 10%) to prevent the influence of rhythm on walking (Schaefer et al., 2015), and a stimulus presentation duration of 500 ms. Each sequence consisted of one third of targets (*i.e.* 90 stimuli for which the participant had to answer "Yes"; Owen et al., 2005) as well as 9 (10%), 13 (14.4%) and 16 (17.8%) lures for the 1-back, 2-back and 3-back, respectively (Szmalec et al., 2011). Each condition lasted 3 minutes.

2.2.3. Kinematic recording

The participants were equipped with 4 mm diameter reflective markers, fixed to specific anatomical landmarks on the body, on top of fitted clothing (Vaughan et al., 1999). The model used for marker placement was an adaption of the Plug-in Gait Full Body (Davis et al., 1991). Only the head, trunk and lower limb markers were used for a total of 27 markers. The three-dimensional positions of the markers were collected at 200 Hz using 12 optoelectronic cameras (VERO 2.2, Vicon Motion Systems Ltd, Oxford, UK). The origin of the benchmark is in the treadmill's centre. Any gap in the raw recorded motion capture data were filled using Vicon Nexus 2.8.1 (Oxford Metrics, Oxford, UK). The 3D marker trajectories were then imported into MATLAB® R2020a (The MathWorks Inc, Natick, Massachusetts, USA) for further analyses.

2.2.4. Electromyographic recording

Once the skin prepared (*i.e.* shaving, light abrasion, cleaning with alcohol solution), surface electromyographic electrodes (EMG Trigno Snap Lead Sensors, Delsys, Inc.) were placed on the *soleus* (Sol) and *gluteus medius* (Gmed) muscles of the dominant lower limb (conversion A/D, 1000 Hz, 10 V). Electrodes were placed longitudinally with respect to the arrangement of muscle fibres (De Luca, 1997) and located according to the recommendations from Surface EMG for Non-Invasive Assessment of Muscles (SENIAM; Hermens et al., 2000).

2.3. Data analysis

2.3.1. Subjective mental workload analysis

In all conditions, the subjective mental workload was estimated using the Raw TLX (R-TLX), that is the sum of the scores of each sub-scale (Hart, 2006).

2.3.2. Cognitive task analysis

The response time (RT in ms) and d prime (d') (Pelegrina et al., 2015; Kumar et al., 2021; Michail et al., 2021) were used to assess cognitive task performance. The d' is the difference between the Z-score obtained from hit rate, Z_{Hit}, and that obtained from false alarms (FA) rate, Z_{FA} (d' = Z_{Hit} - Z_{FA}; Haatveit et al., 2010). The higher the hit rate and the lower the FA, the better the participant's ability to discriminate target and non-target letters. In the present experiment, a d' of 4.5 indicates 100% correct responses and no false alarms.

2.3.3. Kinematic analysis of gait

The kinematic data was low-pass filtered using a 4th order Butterworth zero-phase filter with a cut-off frequency of 6 Hz. Heel strikes were computed from the heel marker trajectory using the algorithm of Zeni et al. (2008). Then, the step velocity (V), step width (W) and lateral body position (z_B, which corresponds to the distance between the centre of the treadmill and the midpoint between the two feet) were extracted, as previously done (Dingwell and Cusumano, 2019; Kazanski et al., 2020). The mean (steadiness), standard deviation (variability) and statistical persistence/anti-persistence of the time series (complexity) of these parameters were then computed. To quantify how stepping fluctuations were regulated from one step to another, the Detrended Fluctuations Analysis (DFA) scaling exponent (α), which assesses the presence and strength of statistical persistence in the investigated time series, was used (Peng et al., 1993; Hausdorff et al., 1997; Damouras et al., 2010; Dingwell et al., 2010). An $\alpha < 0.5$ suggests that the time series contains anti-persistent correlations, *i.e.* subsequent deviations are more likely to be in the opposite direction, consistent with a tightly controlled process. An $\alpha = 0.5$ indicate uncorrelated deviations attributed to noise while $\alpha \approx 0.5$ are typically exhibited by variables that are more tightly regulated. Finally, fluctuations with $\alpha >> 0.5$ (high persistence) are presented by variables that are less tightly regulated (Dingwell and Cusumano, 2010, 2019 ; Kazanski et al., 2020). Given the length of our time series (290 steps), the scaling exponent (α) was computed following the recommendations of Phinyomark et al. (2020). Also, we made a direct control analysis of step-to-step corrections to complete the DFA (Dingwell and Cusumano, 2015). This analysis quantifies how step-to-step corrections of fluctuations might depend nonlinearly on the initial magnitudes of those fluctuations. We directly quantified the degree to which deviations, $q'_n = q_n - \bar{q}$, from the mean value, $\bar{q}$, of a given time series were corrected on the subsequent step by corresponding changes, $\Delta q_{n+1} = q_{n+1} - q_n$, in the opposite direction. From plots of Δq_{n+1} vs. q'_n, we computed the linear slopes (using least-squares regression) and coefficient of determination (R^2) for each corresponding relationship. Parameters tightly controlled would give slopes (M) close to -1 and a higher R^2 (Dingwell and Cusumano, 2019; Kazanski et al., 2020).

2.3.4. EMG analysis of gait

The raw EMG signals were band-pass filtered using a 4th order IIR Butterworth zero-phase filter with cut-off frequencies of 50 and 450 Hz, then rectified and smoothed (low-pass filter) using a 4th order IIR Butterworth zero-phase filter with a cut-off frequency of 10 Hz (Martino et al., 2015). After subtracting the minimum, the muscle activation profiles were normalised in amplitude to the average peak activation of the selected gait cycles (*i.e.* n= 80). Finally, each gait cycle was time-normalised to 200 points by a spline interpolation. In order to assess neuromuscular intra-individual variability (*i.e.* inter-cycle variability within each condition), the variance ratio (VR) was computed for the *soleus* and *gluteus medius* (Hershler and Milner, 1978; Kadaba et al., 1985). The full width at half maximum (FWHM) was used to characterise differences in the duration of muscle activation (Cappellini et al., 2006; Martino et al., 2015). Specifically, cycle-by-cycle, this parameter determines the number of points (percentage of cycle) exceeding half of the maximum activation of each cycle. Then, the values obtained for each of the cycles were averaged to have one value of FWHM per condition (Santuz et al., 2020a). We excluded data of 4 participants from EMG analyses due to the signal quality (*i.e.* movement-related artefacts). Therefore, EMG data from 20 right-handed participants were retained (21.80 ± 2.42 years; BMI: 21.60 ± 2.65).

2.4. Statistical analysis

First, the sphericity assumption was tested was checked using the Mauchly's test. In case of a violation of sphericity, the p-value was adjusted according to the Huynh-Feldt correction (Haverkamp and Beauducel, 2017). A *Task* (1-back, 2-back, 3-back) × *Cognitive condition* (STC, DTC-normal OF, DTC-perturbed OF) repeated-measures analysis of variance (ANOVA) was performed to analyse cognitive (d' and RT) and subjective (R-TLX) variables. The performance achieved during the second single-task walking conditions (*i.e.* performed at the end of the walking blocks) were not considered in the analysis after having observed and statistically confirmed a disengagement or relaxation of the participants during these conditions (Fig. S1, *supplementary materials*). Thus, *Walking condition* (STW, DTW_1-back, DTW_2-back, DTW_3-back) × *Optic flow* (normal OF, perturbed OF) repeated-measures ANOVAs were performed on kinematic (mean, standard deviation, α, M and R^2 for V, W, z_B) and EMG (VR and FWHM for *soleus* and *gluteus medius*) variables. Significance threshold was set at 0.05. Effect sizes were also examined and reported with eta squared ($\eta 2$) or Cohen's d. The statistical analyses were conducted using JASP (version 0.16.1, JASP Team 2020).

3. Results

3.1. Cognitive performance and subjective mental workload

A significant main effect of *Task* was found for both d' ($F(1.59,46.00) = 208.66$; $p_{aj} < 0.001$; $\eta^2 = 0.90$) and RT ($F(2,46) = 81.03$; $p < 0.001$; $\eta^2 = 0.78$). Post-hoc tests revealed a decrease of d' (Figure 3A) and an increase of RT (Figure 3B) with increasing working memory load. Indeed, 3-back performance was significantly lower than 1-back ($p < 0.001$) and 2-back ($p < .001$) performances, and 2-back performance was also significantly lower than 1-back performance ($p < .001$).

A significant main effect of *Task* ($F(2,46) = 178.97$; $p < 0.001$; $\eta^2 = 0.89$) and of *Cognitive condition* ($F(2,46) = 7.77$; $p = 0.001$; $\eta^2 = 0.25$) was found for R-TLX score. Firstly, post-hoc tests revealed an increase in perceived mental demand with increasing working memory load. Secondly, they indicated an increase in perceived mental demand in DTC-perturbed OF conditions compared to both STC ($p = 0.001$) and DTC-normal OF conditions ($p = 0.017$) (Figure 3C).

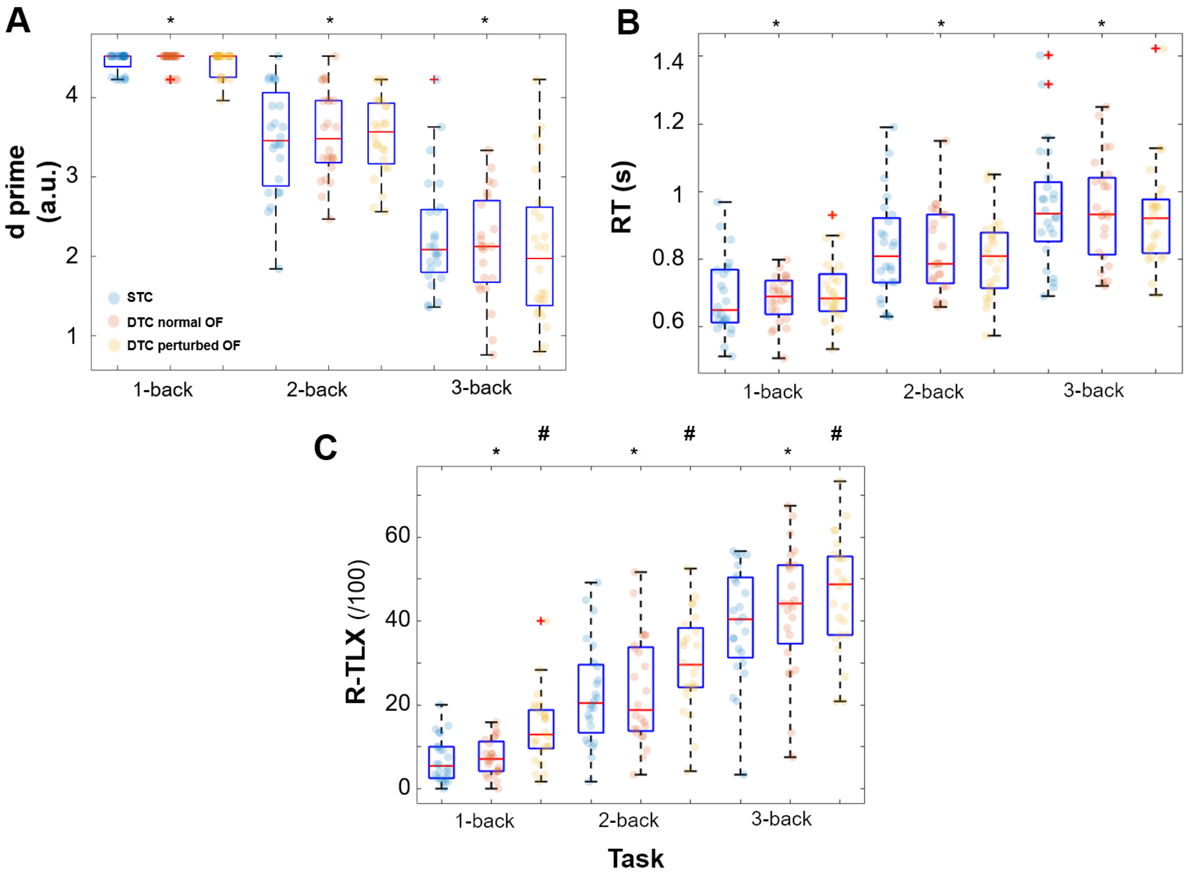

Figure 3. Cognitive performance: A) d', B) RT and C) NASA-TLX obtained in three N-back tasks (1-back, 2-back, 3-back) for three experimental conditions: single task (blue circle) vs dual task walking with normal optical flow (red circle) vs dual task walking with perturbed optical flow (yellow circle). * indicates significant difference between all task's difficulties. # indicates significant difference between the perturbed conditions and others conditions.

3.2. Kinematics of gait

3.2.1. Gait steadiness

A significant main effect of *Walking condition* for mean step velocity ($F(3,69) = 6.03$; $p = 0.001$; $\eta^2 = 0.21$) was found. Post-hoc tests indicated that mean step velocity during DTW-1-back was higher than during both STW and DTW-3-back. Also, a significant main effect of *Optic flow* was found for mean step width ($F(1,23) = 24.45$; $p < 0.001$; $\eta^2 = 0.52$) indicating that mean step width was higher in perturbed OF than in normal OF conditions. An *Optic flow* × *Walking condition* interaction effect was found for mean step width ($F(3,69) = 3.82$; $p = 0.014$; $\eta^2 = 0.14$). Post-hoc tests show no relevant main effect (Figure 4A).

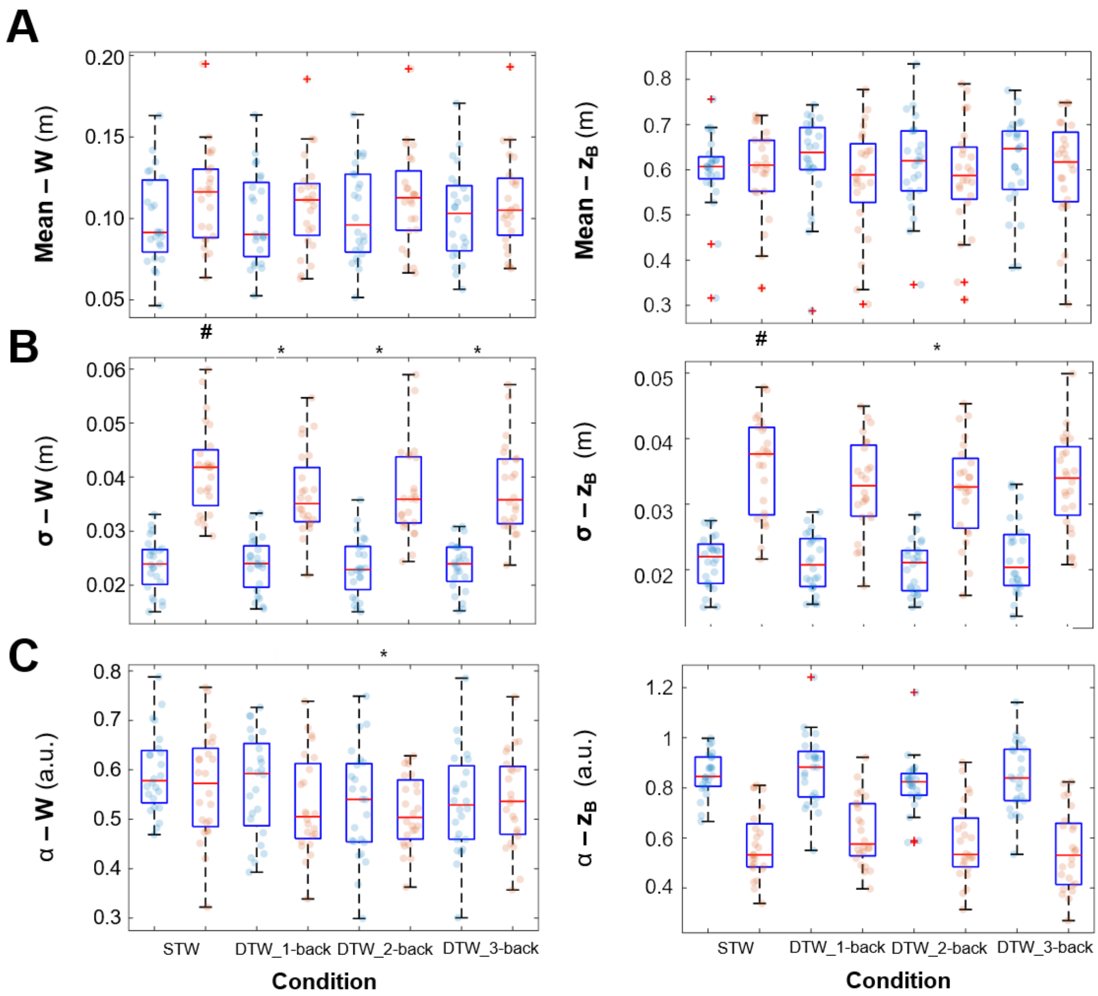

Figure 4. A) Mean, B) Standard deviations (σ) and C) DFA scaling exponents (α) of step width (w, left) and lateral position (z_B, right) for both experimental conditions: normal (blue circle) vs. perturbed optical flow (red circle). * indicates conditions significantly different from the single task. # indicates significant flow x condition interaction effect.

3.2.2. Gait variability (SD: standard deviation)

Significant main effects of *Optic flow* and of *Walking condition* and a significant interaction were found for step velocity SD ($F(1,23) = 24.01$; $p < 0.001$; $\eta^2 = 0.51$ / $F(2.097, 48.232) = 17.54$; $p_{aj} < 0.001$; $\eta^2 = 0.43$ / $F(2.117, 48.701) = 4.18$; $p_{aj} = 0.019$; $\eta^2 = 0.15$), step width SD ($F(1,23) = 118.35$; $p < 0.001$; $\eta^2 = 0.84$ / $F(3,69) = 5.64$; $p = 0.002$; $\eta^2 = 0.20$ / $F(3,69) = 6.96$; $p < 0.001$; $\eta^2 = 0.23$) and z_B SD ($F(1,23) = 75.03$; $p < 0.001$; $\eta^2 = 0.77$ / $F(3,69) = 3.09$; $p = 0.033$; $\eta^2 = 0.12$ / $F(3,69) = 3.32$; $p = 0.025$; $\eta^2 = 0.52$). For the main effects, post-hoc tests indicated significant higher SD values in perturbed OF conditions and lower SD values in DTW conditions, compared to STW conditions. For the interaction effect, they revealed that the effect that perturbed OF had on SD values was attenuated in DTW conditions compared to STW conditions, independently of working memory load (Figure 4B and Figure 5A).

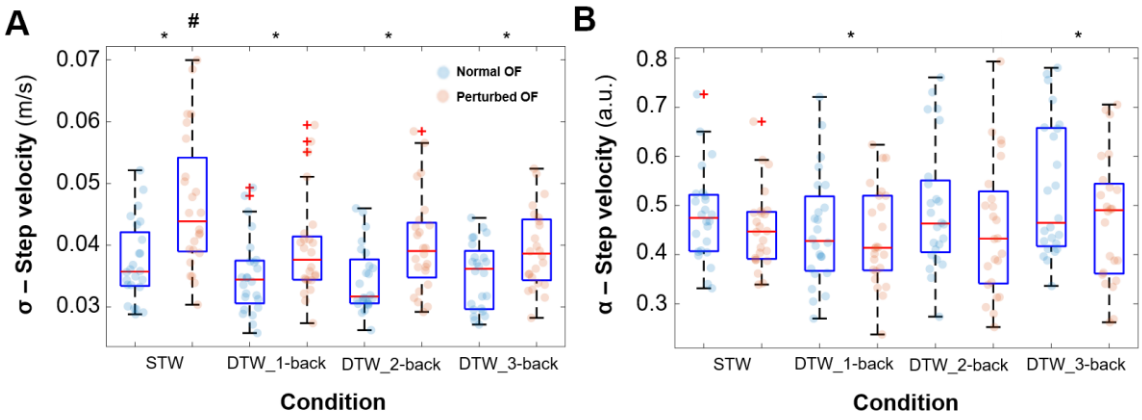

Figure 5. A) Standard deviations (σ) and B) DFA scaling exponents (α) of step velocity for both experimental conditions: normal (blue circle) vs. perturbed optical flow (red circle). * indicates significant difference between conditions. # indicates significant flow x condition interaction effect.

3.2.3. Gait complexity (α: alpha exponent and step-by-step error correction; M & R²)

Significant main effects of *Optic flow* ($F(1,23) = 6.48$; $p = 0.018$; $\eta^2 = 0.22$) and of *Walking condition* ($F(2.448,56.296) = 3.03$; $p_{aj} = 0.046$; $\eta^2 = 0.12$) were found for step velocity α. Post-hoc tests revealed an increase in step velocity anti-persistence in perturbed OF conditions ($\alpha = 0.45$) compared to normal OF conditions ($\alpha = 0.49$) and a decrease in step velocity anti-persistence in DTW-3-back conditions ($\alpha = 0.50$) compared to DTW-1-back conditions ($\alpha = 0.44$, Figure 5B). A significant main effect of *Walking condition* ($F(3,69) = 3.87$; $p = 0.013$; $\eta^2 = 0.14$) was found for step width α. Post-hoc tests revealed a decrease in step width persistence in DTW-2-back conditions ($\alpha = 0.52$) compared to STW conditions ($\alpha = 0.58$, $p = 0.01$). Besides, step width persistence tended to decrease in DTW-3-back conditions ($\alpha = 0.54$, $p = 0.08$) compared to STW conditions (Figure 4C). A significant main effect of *Optic flow* was found for z_B α ($F(1,23) = 71.56$; $p < 0.001$; $\eta^2 = 0.76$) and both W and z_B M (respectively, $F(1,23) = 19.587$; $p < .001$; $\eta^2 = 0.460$ and $F(1,23) = 27.664$; $p < .001$; $\eta^2 = 0.546$) and R^2 (respectively, $F(1,23) = 19.356$; $p < .001$; $\eta^2 = 0.457$ and $F(1,23) = 28.117$; $p < .001$; $\eta^2 = 0.550$). Post-hoc tests revealed a decrease in z_B persistence and steeper correction slope in perturbed OF conditions ($\alpha = 0.58$; $M = -0.29$ and $R^2 = 0.14$) compared to normal OF conditions ($\alpha = 0.85$; $M = -0.23$ and $R^2 = 0.11$). Also, post-hoc tests showed that W correction slopes became more negative in perturbed OF conditions ($M = -0.88$ and $R^2 = 0.44$) compared to normal OF conditions ($M = -0.73$ and $R^2 = 0.36$, Figure 6).

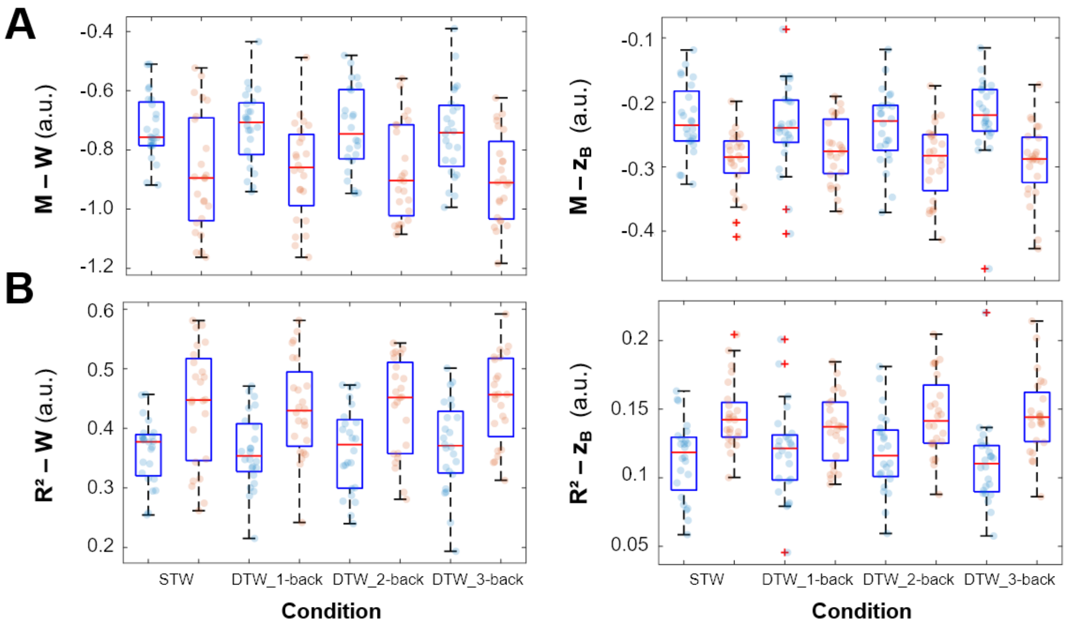

Figure 6. A) Slope of least-squares fits (M) and B) coefficient of determination (R^2) of step width (w, left) and lateral position (z_B, right) for both experimental conditions: normal (blue circle) vs. perturbed optical flow (red circle). * indicates conditions significantly different from the single task. # indicates significant flow x condition interaction effect.

All statistical results related to gait kinematics and descriptive results are reported respectively in Tables S1 and S2 and Tables S3 and S4 (Supplementary materials).

3.3. EMG of gait

Significant main effects of *Optic flow* and *Walking condition* for VR of both *soleus* muscle ($F(1,19) = 16.46$; $p < 0.001$; $\eta^2 = 0.46$ / $F(2.111,40.109) = 20.18$; $p_{aj} < 0.001$; $\eta^2 = 0.52$) and *gluteus medius* muscle ($F(1,19) = 33.03$; $p < 0.001$; $\eta^2 = 0.64$ / $F(3,57) = 17.11$; $p < 0.001$; $\eta^2 = 0.47$). Post-hoc tests indicated higher VR values for both muscles in perturbed OF conditions compared to normal OF conditions, and lower VR values in DTW conditions compared to STW conditions (Figure 7A). A significant main effect of *Walking condition* ($F(3,57) = 4.05$; $p = 0.011$; $\eta^2 = 0.17$) was found for FWHM of the *soleus* muscle. Post-hoc results indicated lower FWHM values in STW conditions compared to both DTW-1-back and DTW-2-back conditions (Figure 7B).

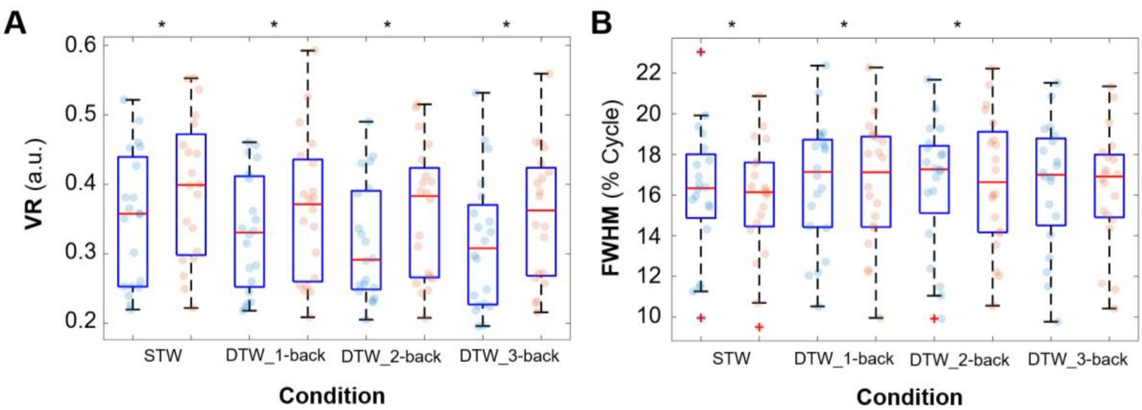

Figure 7. A) Variance ratio of *gluteus medius* and B) Full width at half maximum of *soleus* for both experimental conditions: normal (blue circle) vs. perturbed optical flow (red circle). * indicates conditions significantly different from the single task.

4. Discussion

The aim of the present study was to assess the attentional cost associated with gait control under ML optic flow perturbations in young adults. In line with our first hypothesis, our results confirmed that attentional sharing lowered the negative effects of these perturbations on gait control at both the kinematic and neuromuscular levels due to fewer attentional resources allocated to processing of visual inputs; part of these limited resources being redirected to the concurrent auditory N-back task to walking. However, contrary to our second hypothesis, an increase in concurrent working memory did not amplify this phenomenon.

As a main result of this study, we found that sharing attention decreased the negative effect of perturbed OF on gait as observed by Malcolm et al. (2018) on the standard deviation of the head position in AP and ML directions. Indeed, in OF perturbations, we observed an increase in step-to-step variability in all gait parameters (see also Supplementary material; Stokes et al., 2017; Kazanski et al., 2020; Osaba et al., 2020; Selgrade et al., 2020; Dingwell et al., 2021; Song et al., 2022) and a decrease in the dual task condition. Specifically, this increase in perturbed OF is reduced in dual-task condition independently of working memory load. Thus, this finding suggests that the impact of destabilizing OF perturbations is modulated when attention is diverted toward auditory cues to process a secondary task. One explanation may rely on the sensory modality of the cognitive task: a visual task would have increased engagement in the visual field, whereas an auditory task prompted participants to redirect their attention elsewhere, which has resulted in a reduction of the impact of OF perturbations on gait control (Wickens, 2002; Lavie et al., 2004; Redfern et al., 2017; Pechtl et al., 2020; Brockhoff et al., 2022). Also, it is consistent with the idea that resolving sensory conflict requires executive control of attention (Yogev-Seligmann et al., 2008; Redfern et al., 2017). Indeed, adding an auditory working memory task decreased the processing (perception) of perturbed optic flow whatever the task difficulty.

Theses interpretations are consistent with observed cognitive performance. Similarly to Pechtl et al. (2020), the performance in the auditory N-back tasks (i.e., d' and TR) did not differ between sitting and walking conditions, so the young adult participants of this study were able to sustain their cognitive engagement regardless of the context. However, participants' perceived mental load (i.e., R-TLX score) was significantly higher under walking with perturbed OF compared to walking with normal OF and sitting conditions. Thus, walking under perturbed OF was perceived as the most challenging, requiring a great deal of concentration and mental effort to handle both tasks at hand. The fact that the cognitive performance remained practically equivalent under both single- and dual-task conditions suggests that the same amount of attentional resources were used in both conditions; i.e. a prioritisation of the cognitive task. This was in contrast to a prioritisation of the walking task found by Kao and Pierro (2022) and might be explained by the modality of the perturbation (visual vs. mechanical, Kazanski et al., 2020).

The effects of OF perturbations on variability (standard deviation) and, also, gait steadiness (mean) were consistent with those reported previously. Indeed, young adults walked with wider steps to improve their ML balance during walking under perturbed OF (Beurskens et al., 2014; Kao and Pierro, 2022). This was in line with model simulations suggesting that walking with wide steps decreased the control precision required to maintain balance (Kuo, 1999), and with experimental studies providing indirect evidence that humans walked with more laterally placed steps (Rankin et al., 2014) and more metabolically costly gait patterns (Monsch et al., 2012) when they perceived a challenge to their balance. Hence, walking with wider steps is likely a beneficial adaptation to counteract the increase in step-to-step variability induced by the OF perturbations in order to maintain their stability (Kazanski et al., 2020). More broadly, these findings confirmed the prominent role of vision in the control of balance during walking, particularly in the identification of self-motion, which is an important source of information for navigation (Logan et al., 2010).

Furthermore, we observed that fluctuations in step velocity exhibited an increase in antipersistence during perturbed OF supporting the importance of considering this parameter when walking on treadmill at fixed speed (Dingwell et al., 2010; Decker et al., 2013; Terrier and Dériaz, 2013; Roerdink et al., 2015). Direct control analysis confirmed that the step-by-step fluctuations were strongly controlled (M > -1.0). However, we did not observe any modification of this regulation under perturbed optic flow; the walk being constrained by the treadmill whatever the flow (Hausdorff et al., 1997; Dingwell et al., 2010).

In line with literature (Kazanski et al., 2020; Render et al., 2021), results confirmed that under perturbed conditions, young adults tightened their ML stepping regulation. Indeed, they increased step-to-step corrections of lateral position and step width (closer to -1.0), decreased persistence of the fluctuations for z_B and maintained that for the step width. Under free single-task walking, young adults prioritised step width control (α = 0.59) over lateral position control (α = 0.85). These results were in line with the stepping regulation simulations from Dingwell et al. (2019). Thus, in perturbed condition, step width regulation was maintained because it is already highly persistent (controlled) in normal condition contrary to zB regulation. Also, step width exhibited steeper correction slopes (M = -0.72) and a higher R^2 (R^2 = 0.36) compared to the lateral position (respectively, M = -0.28 and R^2 = 0.11), reflecting stronger corrections for W than for z_B (Dingwell and Cusumano, 2019; Kazanski et al., 2020). Overall, this agreed with multi-objective control simulations (Dingwell et Cusumano., 2019), suggesting that participants increased their control to enact greater corrections for deviations in both W and z_B. Therefore, this increased regulation might allocate more attentional resources at the expense of the processing of other parallel tasks (Boisgontier et al., 2011).

The increased variability under perturbed OF is also visible at the neuromuscular level as we found an increased variance ratio for both the *gluteus medius* and *soleus* muscles. To our knowledge, it is the first demonstration of this effect in the literature. This finding is complementary to those of Stokes et al. (2017), who observed an increase in lateral foot placement magnitude positively correlated with that of *gluteus medius* muscle activation in response to OF perturbations. On the other hand, contrary to what was expected, no difference was observed in the FWHM of both muscles under OF perturbations. This discrepancy may arise from the types of perturbation. Indeed, Martino et al. (2015) and Santuz et al. (2020b) investigated walking under mechanical disturbances (respectively, on a slippery floor or beam, an uneven surface, or support displacements), while optic flow perturbations were entirely perceptual. In considering that young adults relied less on co-activation than older to maintain the equilibrium Acuña et al (2019), the absence of OF perturbation effect on the FWHM of both muscles suggested that participants manage to inhibit (i.e. not perceive) partially the disturbance which is not possible with mechanical perturbation.

As expected, under both single-task and dual-task conditions, cognitive performance decreased (greater TR and lower d'; Pelegrina et al., 2015; Lamichhane et al., 2020) and mental effort increased (greater R-TLX score; Sadeghian et al., 2021) with increasing working memory load, thus validating our incremental dual-task paradigm (3-back > 2-back > 1-back). Also, it should be pointed out that the performance in the 3-back task was, on average across participants and conditions , reduced by half (d' ≈ 2.26), indicating that participants failed to perform this task correctly. Pelegrina et al. (2015) defined performance failure when the hit rate is less than 60%, which was the case in 29% of the 3-back tasks performed in the present study. In these cases, we cannot define whether the participants' dual-task performance can really be attributed to attentional sharing or to disengagement due to an overly complex task (Ruffieux et al., 2015; Doherty et al., 2019). To overcome this, it would have been possible to set up a titration procedure, by adjusting the difficulty of each n-task to the working memory capacity of each participant (Hunter et al., 2018; Doherty et al., 2019; Jaroslawska et al., 2021). However, Stern (2009) argued that this approach does not really solve the problem because participants might be matched on one measure, such as performance accuracy, but still differed on another measure, such as response time. Furthermore, the R-TLX score results confirm the engagement in the 3-back task across all conditions.

At the kinematic level, participants adopted a more cautious and rigid gait pattern under dual-task conditions regardless of the difficulty of the simultaneous n-back task. Indeed, participants became less variable under shared attention. These results are in agreement with the literature, reporting a decrease in variability of step width (Grabiner and Troy, 2005; Malcolm et al., 2018), step length and step velocity (Lövdén et al., 2008). Wrightson et al. (2016) proposed that, regardless of the cognitive task type (among which the n-back task), participants may redirect cognitive processes from gait to cognitive tasks. However, we did not observe any effect of the difficulty of this task on steadiness and variability. Thus, in dual task, the system rigidifies the gait regardless of the difficulty of the cognitive task in order to stabilise the system and release attentional resources for the cognitive task.

Only the complexity measures were sensitive enough to underline how the deviations are regulated step by step as a function of load. Indeed, loading cognitive control (executive) processes (here, working memory) decreased anti-persistence in step velocity (V) and persistence in step width (W). Less anti-persistent fluctuations in V reflected an impaired ability to rapidly correct speed deviations on subsequent steps in AP direction. This result is compatible with the idea that anti-persistence in gait results from executive processes (Decker et al., 2013). Thus, when the secondary cognitive task involves the executive functions, the CNS releases part of the attention allocated to the control of step velocity which increasingly relies on automatic mechanisms (Bayot et al., 2018). Also, we did not observe any modification of the correction slopes and the associated R^2. Indeed, the tightened control is maintained whatever the conditions of the walking task in order stay within the limits of the treadmill (fixed speed). The treadmill constraint is the main factor/cause in the regulation of the step speed.

Fluctuations in step width showed a loss of persistence when an executive cognitive task was performed in parallel, indicating that more resources were focused on its regulation. It suggested that, under dual-task constraints, participants may tend prioritise control in ML according to the OF perturbations, over control in AP, which would explain the maintenance of persistence in z_B and decrease in W. Thus, participants cannot afford to release their attention (and to function automatically) in favour of the cognitive task as walking is more unstable in ML (Bauby and Kuo, 2000). In addition, we noticed an increase in step width regulation in DTW-2b compared to STW but not in DTW-3b. Indeed, in the 2-back task, participants were able to share their resources between the two tasks by increasing step width regulation, which was not the case in 3-back. In fact, in the DTW-3b task, they maintained a sufficient level of regulation to perform both tasks efficiently by decreasing control in the AP.

At the neurophysiological level, the increased activation time of the *soleus* muscle (FWHM) suggested that gait was less stable under dual-task than single-task walking. Therefore, as previously showed (Martino et al., 2015), the CNS tended to widen muscle activation time to cope with this dynamic instability. Considering the increased VR with dual-task, all of these neuromuscular changes support that sharing attention led to a more rigid gait control, as observed at the kinematic level in order to stabilise locomotion (Verrel et al., 2009; Kao and Pierro, 2022).

The fact that the cognitive performance remained practically equivalent under both single- and dual-task conditions highlights the task prioritisation strategy used by the participants. Indeed, a decreased gait control combined with a preserved cognitive performance under dual-task conditions reflects a “posture second” strategy, where the cognitive task, and thus the attentional resources allocated to it, is prioritised over the gait task (Shumway-Cook et al., 1997; Bloem et al., 2006; Yogev-Seligmann et al., 2012a, 2012b; Plummer et al., 2013). We observed a decrease in the control of step velocity fluctuations as the cognitive load increased in line with this strategy (Small et al., 2021). However, we noticed an increase in step width regulation in DTW-2b and a maintain in DTW-3b. In fact, in DTW-3b, they partially disengaged from locomotion (especially in AP control) to allocate most of their attention to 3-back and perform both tasks efficiently. Thus, ML regulation does not prevent the adoption by the participants of “posture second” strategy.

There as some limitations in this study that should be mentioned. First, single-task conditions were always performed before the dual-tasks since we did not retain from the analyses those performed after the dual tasks, even if substantial familiarization had taken place beforehand. Further analysis will thus be conducted to understand the differences in both single tasks. Also, participants walked at a fixed speed on a treadmill. The differences between such modality and overground walking (i.e., everyday walking) have been extensively studied and highlighted (Dingwell et al., 2001; Wrightson et al., 2020). Walking on treadmill, even if less ecological, allows the possibility to collect more walking cycles under standardised conditions and the use of a more complex dual-task paradigm. An intermediate approach could be to use the treadmill with the self-paced mode (Yang and King, 2016; Choi et al., 2017; Kao and Pierro, 2022) which we plan to do in future studies. To better understand the cognitive mechanisms related to sensory integration during walking, it will be necessary, in the future, to perform this type of protocol on elderly (asymptomatic) people or patients with central or peripheral (i.e. sensory) deficits.

To conclude, the present study highlighted the attentional cost of optic flow perturbation integration as performing a secondary auditory task mitigated the effects of this perturbation at both the kinematic and muscular variability parameters. Taken together, these results confirm that the processing of incongruent visual signals is attentionally, not just executive, costly. Also, the non-linear analyses performed on our kinematic variables are the only ones that allowed us to highlight the prioritisation strategies of our participants; respectively, the relaxation and maintenance of the control of the step speed and ML parameters when the task becomes too complex (i.e. highly executive) associated with the maintenance of cognitive performance.

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Supplementary Material

A condition (Normal Flow, Perturbed Flow) x Attempt (1, 2) repeated measures ANOVA showed a significant main effect of condition ($F(1,22) = 35.153$; $p < 0.001$; $\eta^2 = 0.615$), of attempt ($F(1,22) = 29.308$; $p < 0.001$; $\eta^2 = 0.571$) and interaction effect ($F(1,22) = 20.637$; $p < 0.001$; $\eta^2 = 0.484$). Post-hoc tests revealed an increase in perceived mental load with the perturbed optical flow and an important decrease in the second attempt than first attempt. Concerning interaction effect, post-hoc revealed that the second attempt reduce perturbed flow effect. Indeed, the perceived mental load is almost null with normal flow, so the main effects observed primarily concern the perturbed flow condition. We hypothesize that a decrease in vigilance (i.e., cognitive relaxation) at the end of the blocks would have impacted the performance achieved in second attempt of walking single task. Another hypothesis would be a learning effect. In any case, as this was not the aim of the present study, so as not to impact on our analyses, we decided to extract the second single task from our analysis

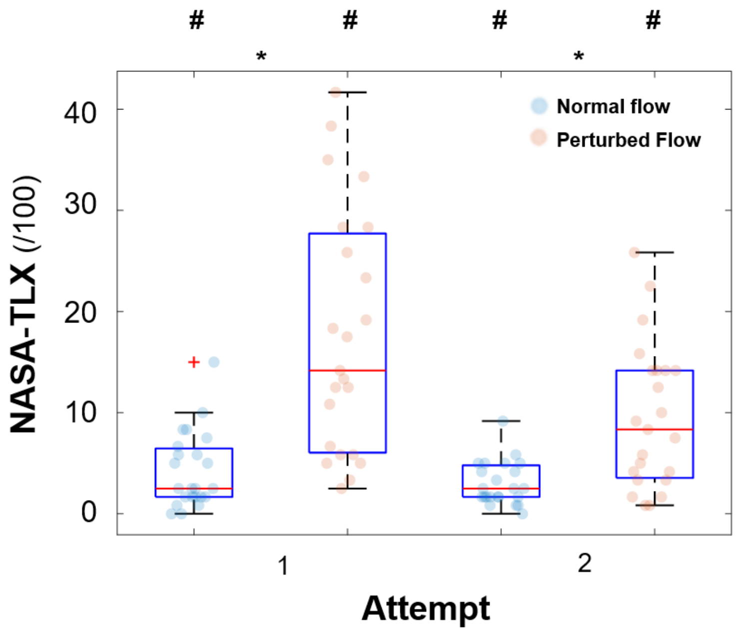

Figure S1. NASA-TLX obtained in different attempt of locomotor single-task (1, 2) for both experimental conditions: normal (blue circle) vs. perturbed optical flow (red circle). * indicates significant difference in both conditions. # indicates significant Attempt x Condition interaction effect. All conditions are different from each other except both conditions in normal flow

Variable		FN			
		ST	DT_1b	DT_2b	DT_3b
Step Width	*Mean (m)*	0.099 ± 0.028	0.100 ± 0.028	0.103 ± 0.029	0.102 ± 0.028
	Std (m)	0.024 ± 0.005	0.024 ± 0.005	0.023 ± 0.005	0.023 ± 0.004
	M (a.u.)	-0.725 ± 0.109	-0.722 ± 0.125	-0.730 ± 0.143	-0.741 ± 0.150
	R^2 (a.u.)	0.362 ± 0.054	0.362 ± 0.062	0.365 ± 0.071	0.371 ± 0.076
	Alpha (a.u.)	0.594 ± 0.082	0.569 ± 0.103	0.531 ± 0.110	0.534 ± 0.104
Zb	*Mean (m)*	-0.020 ± 0.017	-0.016 ± 0.021	-0.017 ± 0.021	-0.017 ± 0.021
	Std (m)	0.021 ± 0.004	0.021 ± 0.004	0.020 ± 0.004	0.022 ± 0.006
	M (a.u.)	-0.227 ± 0.056	-0.238 ± 0.068	-0.239 ± 0.059	-0.221 ± 0.065
	R^2 (a.u.)	0.113 ± 0.028	0.119 ± 0.034	0.118 ± 0.029	0.110 ± 0.031
	Alpha (a.u.)	0.853 ± 0.086	0.871 ± 0.143	0.815 ± 0.114	0.843 ± 0.138
Step Velocity	*Mean (m/s)*	1.117 ± 0.132	1.117 ± 0.132	1.117 ± 0.132	1.117 ± 0.132
	Std (m/s)	0.037 ± 0.006	0.035 ± 0.006	0.034 ± 0.005	0.035 ± 0.005
	M (a.u.)	-1.210 ± 0.228	-1.214 ± 0.201	-1.192 ± 0.220	-1.154 ± 0.186
	R^2 (a.u.)	0.606 ± 0.114	0.607 ± 0.100	0.596 ± 0.109	0.578 ± 0.092
	Alpha (a.u.)	0.481 ± 0.094	0.450 ± 0.114	0.496 ± 0.130	0.531 ± 0.139
Step Length	*Mean (m)*	0.626 ± 0.066	0.622 ± 0.067	0.624 ± 0.065	0.624 ± 0.067
	Std (m/s)	0.016 ± 0.003	0.015 ± 0.003	0.015 ± 0.003	0.015 ± 0.003
Step Time	*Mean (s)*	0.562 ± 0.032	0.559 ± 0.030	0.560 ± 0.031	0.560 ± 0.030
	Std (m/s)	0.012 ± 0.003	0.011 ± 0.003	0.012 ± 0.002	0.012 ± 0.003
Stance phase	*Mean (s)*	0.739 ± 0.048	0.734 ± 0.046	0.736 ± 0.047	0.736 ± 0.046
	Std (m/s)	0.014 ± 0.003	0.014 ± 0.003	0.014 ± 0.003	0.014 ± 0.003
Single Support	*Mean (s)*	0.656 ± 0.011	0.657 ± 0.011	0.657 ± 0.011	0.657 ± 0.010
	Std (m/s)	0.006 ± 9.275e-4	0.006 ± 8.747e-4	0.006 ± 0.001	0.006 ± 0.001
Double Support	*Mean (s)*	0.176 ± 0.019	0.176 ± 0.018	0.176 ± 0.019	0.176 ± 0.018
	Std (m/s)	0.010 ± 0.002	0.009 ± 0.002	0.009 ± 0.002	0.009 ± 0.002
eGVI	*Mean (a.u.)*	106.821 ± 4.948	107.782 ± 5.061	108.280 ± 3.990	108.026 ± 4.766
Walk ratio	*Mean (m/step/s)*	0.006 ± 6.935e-4	0.006 ± 6.932e-4	0.006 ± 6.822e-4	0.006 ± 7.064e-4
Cadence	*Mean (step/min)*	106.947 ± 5.992	107.716 ± 5.818	107.430 ± 5.943	107.368 ± 5.895

***Table S1**. Descriptive data (mean ± standard deviation) from kinematic locomotor parameters for each Condition (single task ; dual task_1-back ; dual task_2-back, dual task_3-back) in normal (A) and perturbed (B) flow.*

Variable		OF			
		ST	**DT_1b**	**DT_2b**	**DT_3b**
Step Width	*Mean (m)*	0.113 ± 0.029	0.109 ± 0.029	0.112 ± 0.030	0.110 ± 0.029
	Std (m)	0.041 ± 0.008	0.037 ± 0.008	0.037 ± 0.009	0.038 ± 0.008
	M (a.u.)	-0.880 ± 0.199	-0.869 ± 0.165	-0.873 ± 0.164	-0.897 ± 0.150
	R^2 (a.u.)	0.440 ± 0.099	0.434 ± 0.083	0.436 ± 0.081	0.449 ± 0.075
	Alpha (a.u.)	0.567 ± 0.108	0.529 ± 0.100	0.514 ± 0.073	0.541 ± 0.096
Zb	*Mean (m)*	-0.020 ± 0.020	-0.024 ± 0.024	-0.023 ± 0.024	-0.020 ± 0.023
	Std (m)	0.036 ± 0.008	0.033 ± 0.007	0.032 ± 0.007	0.033 ± 0.007
	M (a.u.)	-0.291 ± 0.049	-0.274 ± 0.050	-0.291 ± 0.059	-0.292 ± 0.058
	R^2 (a.u.)	0.146 ± 0.025	0.136 ± 0.025	0.145 ± 0.029	0.146 ± 0.029
	Alpha (a.u.)	0.572 ± 0.136	0.624 ± 0.136	0.584 ± 0.154	0.549 ± 0.159
Step Velocity	*Mean (m/s)*	1.116 ± 0.132	1.118 ± 0.132	1.117 ± 0.132	1.117 ± 0.132
	Std (m/s)	0.047 ± 0.011	0.040 ± 0.008	0.040 ± 0.008	0.039 ± 0.006
	M (a.u.)	-1.222 ± 0.156	-1.195 ± 0.186	-1.232 ± 0.219	-1.189 ± 0.207
	R^2 (a.u.)	0.611 ± 0.078	0.597 ± 0.093	0.616 ± 0.109	0.594 ± 0.103
	Alpha (a.u.)	0.453 ± 0.081	0.437 ± 0.100	0.453 ± 0.137	0.472 ± 0.129
Step Length	*Mean (m)*	0.610 ± 0.071	0.612 ± 0.068	0.613 ± 0.069	0.614 ± 0.069
	Std (m/s)	0.023 ± 0.005	0.019 ± 0.005	0.019 ± 0.005	0.018 ± 0.004
Step Time	*Mean (s)*	0.548 ± 0.028	0.549 ± 0.027	0.550 ± 0.028	0.552 ± 0.029
	Std (m/s)	0.016 ± 0.004	0.013 ± 0.003	0.013 ± 0.004	0.013 ± 0.003
Stance phase	*Mean (s)*	0.718 ± 0.043	0.721 ± 0.042	0.722 ± 0.043	0.724 ± 0.044
	Std (m/s)	0.020 ± 0.006	0.017 ± 0.004	0.017 ± 0.005	0.017 ± 0.004
Single Support	*Mean (s)*	0.655 ± 0.011	0.657 ± 0.011	0.656 ± 0.010	0.656 ± 0.010
	Std (m/s)	0.009 ± 0.003	0.007 ± 0.002	0.007 ± 0.002	0.007 ± 0.002
Double Support	*Mean (s)*	0.171 ± 0.017	0.172 ± 0.017	0.172 ± 0.017	0.172 ± 0.017
	Std (m/s)	0.012 ± 0.003	0.010 ± 0.003	0.010 ± 0.003	0.010 ± 0.003
eGVI	*Mean (a.u.)*	107.394 ± 8.526	105.124 ± 5.983	106.359 ± 6.134	105.198 ± 5.817
Walk ratio	*Mean (m/step/s)*	0.006 ± 7.527e-4	0.006 ± 7.014e-4	0.006 ± 7.194e-4	0.006 ± 7.272e-4
Cadence	*Mean (step/min)*	109.801 ± 5.883	109.622 ± 5.644	109.407 ± 5.835	109.088 ± 5.890

***Table S2**. Descriptive data (mean ± standard deviation) from kinematic locomotor parameters for each Condition (single task ; dual task_1-back ; dual task_2-back, dual task_3-back) in perturbed flow.*

Variable		Flux	Condition	Flux x Condition	**Holm's** (Condition effects)	**Holm's** (Flux x Condition effects)
Step Width	*Mean (m)*	$F(1,23) = 24.449$ $p < .001$ $\eta^2 = 0.515$	n.s.	$F(3,69) = 3.821$ $p = .014$ $\eta^2 = 0.142$	n.s.	n/a
	Std (m)	$F(1,23) =$ 118.351 $p < .001$ $\eta^2 = 0.837$	$F(3,69) = 5.637$ $p = .002$ $\eta^2 = 0.197$	$F(3,69) = 6.962$ $p < .001$ $\eta^2 = 0.232$	STW > DTW-1b : p=.006 STW > DTW-2b : p=.003 STW > DTW-3b : p=.024	STW_POF > All
Zb	*Mean (m)*	n.s.	n.s.	n.s.	n.s.	n.s.
	Std (m)	$F(1,23) = 75.032$ $p < .001$ $\eta^2 = 0.765$	$F(3,69) = 3.094$ $p = .033$ $\eta^2 = 0.119$	$F(3,69) = 3.321$ $p = .025$ $\eta^2 = 0.515$	STW > DTW-2b : p=.031	STW_POF > All
Step Velocity	*Mean (m/s)*	n.s.	$F(3, 69) = 6.030$ $p = .001$ $\eta^2 = 0.208$	n.s.	DTW_1b > STW : p<.001 DTW_1b > DTW_3b : p = .021	n.s.
	Std (m/s)	$F(1,23) = 24.014$ $p < .001$ $\eta^2 = 0.511$	$F(2.10, 48.23) = 17.542$ $p_{aj} < .001$ $\eta^2 = 0.433$	$F(2.12,48.70) = 4.178$ $p_{aj} = .019$ $\eta^2 = 0.154$	STW > DTW-1b : p<.001 STW > DTW-2b : p<.001 STW > DTW-3b : p<.001	STW_POF > All
Step Length	*Mean (m)*	$F(1,23) = 57.778$ $p < .001$ $\eta^2 = 0.715$	n.s.	$F(3,69) = 5.618$ $p = .002$ $\eta^2 = 0.196$	n.s.	n/a
	Std (m)	$F(1,23) = 35.661$ $p < .001$ $\eta^2 = 0.608$	$F(2.04,46.88) = 20.697$ $paj < .001$ $\eta^2 = 0.474$	$F(2.18,50.13) = 7.022$ $p_{aj} = .002$ $\eta^2 = 0.234$	STW > DTW-1b : p<.001 STW > DTW-2b : p<.001 STW > DTW-3b : p<.001	STW_POF > All
Step Time	*Mean (s)*	$F(1,23) = 54.093$ $p < .001$ $n^2 = 0.702$	n.s.	$F(3,69) = 5.216$ $p = .003$ $n^2 = 0.185$	n.s.	n/a
	Std (s)	$F(1,23) = 20.995$ $p < .001$ $\eta^2 = 0.477$	$F(2.37,54.57) = 20.195$ $p_{aj} < .001$ $\eta^2 = 0.468$	$F(3,69) = 7.022$ $p < .001$ $\eta^2 = 0338$	STW > DTW-1b : p<.001 STW > DTW-2b : p<.001 STW > DTW-3b : p<.001	STW_POF > All

Stance phase	*Mean (s)*	F(1,23) = 54.523 $p < .001$ $\eta^2 = 0.703$	n.s.	F(3,69) = 4.625 $p = .005$ $\eta^2 = 0.167$	n.s.	n/a
	Std (s)	F(1,23) = 37.957 $p < .001$ $\eta^2 = 0.623$	F(3,69) = 25.057 $p < .001$ $\eta^2 = 0.521$	F(2.28,52,34) = 13.005 $p_{aj} < .001$ $\eta^2 = 0.361$	STW > DTW-1b : p<.001 STW > DTW-2b : p<.001 STW > DTW-3b : p<.001	STW_POF > All
Single Support	*Mean (s)*	F(1,23) = 4.786 $p = .039$ $\eta^2 = 0.172$	F(1.85,42.58)=5.455 $p_{aj} = .009$ $\eta^2 = 0.192$	n.s.	STW < DTW-1b : p=.001 STW < DTW-2b : p=,052	n.s.
	Std (s)	F(1,23) = 44.555 $p < .001$ $\eta^2 = 0.660$	F(3,69) = 21.476 $p < .001$ $\eta^2 = 0.483$	F(2.35,54.07)=8.671 $p_{aj} < .001$ $\eta^2 = 0.274$	STW > DTW-1b : p<.001 STW > DTW-2b : p<.001 STW > DTW-3b : p<.001	STW_POF > All
Double Support	*Mean (s)*	F(1,23) = 40.621 $p < .001$ $n^2 = 0.702$	n.s.	n.s.	n.s.	n.s.
	Std (s)	F(1,23) = 16.195 $p < .001$ $\eta^2 = 0.413$	F(3,69) = 20.422 $p < .001$ $\eta^2 = 0.470$	F(3,69) = 6.732 $p < .001$ $\eta^2 = 0.226$	STW > DTW-1b : p<.001 STW > DTW-2b : p<.001 STW > DTW-3b : p<.001	STW_POF > All
eGVI	*Mean (a.u.)*	n.s.	n.s.	F(2.31,53.10)=3.701 $p_{aj} = .026$ $\eta^2 = 0.139$	n.s.	n.s.
Walk ratio	*Mean (m/step/s)*	F(1,23) = 58.694 $p < .001$ $n^2 = 0.718$	n.s.	F(3,69) = 5.519 $p = .002$ $n^2 = 0.194$	n.s.	n/a
Cadence	*Mean (step/min)*	F(1,23) = 54.947 $p < .001$ $n^2 = 0.705$	n.s.	F(3,69) = 5.303 $p = .002$ $n^2 = 0.187$	n.s.	n/a

Table S3*. Statistical results for variances analysis between Condition (single task ; dual task_1-back ; dual task_2-back, dual task_3-back) and Flow (normal, perturbed) for steadiness and variability kinematic locomotor parameters.*

Variable		Flux	Condition	Flux x Condition	**Holm's** (Condition effects)	**Holm's** (Flux x Condition effects)
Step Width	*M (a.u.)*	$F(1,23) = 19.587$ $p < .001$ $\eta^2 = 0.460$	n.s.	n.s.	n.s.	n.s.
	R^2 (a.u.)	$F(1,23) = 19.356$ $p < .001$ $\eta^2 = 0.457$	n.s.	n.s.	n.s.	n.s.
	Alpha (a.u.)	n.s.	$F(3,69) = 3.873$ $p = .013$ $\eta^2 = 0.144$	n.s.	STW < DTW-2b : p=.010	n.s.
Zb	*M (a.u.)*	$F(1,23) = 27.664$ $p < .001$ $\eta^2 = 0.546$	n.s.	n.s.	n.s.	n.s.
	R^2 (a.u.)	$F(1,23) = 28.117$ $p < .001$ $\eta^2 = 0.550$	n.s.	n.s.	n.s.	n.s.
	Alpha (a.u.)	$F(1,23) = 71.556$ $p < .001$ $\eta^2 = 0.757$	n.s.	n.s.	n.s.	n.s.
Step Velocity	*M (a.u.)*	n.s.	n.s.	n.s.	n.s.	n.s.
	R^2 (a.u.)	n.s.	n.s.	n.s.	n.s.	n.s.
	Alpha (a.u.)	$F(1,23) = 6.483$ $p = .018$ $\eta^2 = 0.220$	$F(2.45,56.30) = 3.034$ $p_{aj} = .046$ $\eta^2 = 0.117$	n.s.	DTW-3b > DTW-1b : p = .023	n.s.

Table S4. *Statistical results for variances analysis between Condition (single task ; dual task_1-back ; dual task_2-back, dual task_3-back) and Flow (normal, perturbed) for complexity kinematic locomotor parameters.*