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Dynamic Asymmetries Do Not Match Spatiotemporal Step Asymmetries during Split-Belt Walking

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Abstract: While walking on split-belt treadmills (two belts running at different speeds), the slower limb shows longer anterior steps than the limb dragged by the faster belt. After returning to basal conditions, the step length asymmetry is transiently reversed (after-effect). The lower limb joint dynamics, however, were not thoroughly investigated. In this study, 12 healthy adults walked on a force-sensorised split-belt treadmill for 15 min. Belts rotated at 0.4 m s^{-1} on both sides, or 0.4 and 1.2 m s^{-1} under the non-dominant and dominant legs, respectively. Spatiotemporal step parameters, ankle power and work, and the actual mean velocity of the body's centre of mass (CoM) were computed. On the faster side, ankle power and work increased, while step length and stance time decreased. The mean velocity of the CoM slightly decreased. As an after-effect, modest converse asymmetries developed, fading within 2–5 min. These results may help to decide which belt should be assigned to the paretic and the unaffected lower limb when split-belt walking is applied for rehabilitation research in hemiparesis.

Keywords: split-belt treadmill; walking; asymmetry; rehabilitation

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1. Introduction

Walking on a split-belt treadmill (two belts running at different velocities) was proposed both as an experimental model for studying the neural control of walking [1] and as a form of "re-symmetrising" therapeutic exercise for patients with unilateral motor impairments [2–6]. The split-walking paradigm implies spatiotemporal, kinematic, and dynamic asymmetries. In healthy participants, the faster leg is passively dragged backwards by the corresponding belt relative to the slower leg and is forced to make up for the lost space. Therefore, the step length (SL) and stance time are shortened on the faster belt, giving rise to the appearance of a pathologic-like "escape limp", wherein the plantar flexor muscles, which are the main drivers of body propulsion [7–11], are forced to provide more power and work to sustain the "escape" [7]. Humans adapt within a few strides to this unusual locomotory pattern. On return to baseline after 5–20 min ("tied" belts, post-adaptation), an "after-effect" was described [12–15], i.e., a transient reversal of the kinematic asymmetries. The after-effect, which is a common phenomenon in various human behaviours, is considered here as evidence of learning and transient retention of the "split" motor pattern [15–17] and can reduce the asymmetries in SL [15,18], characterising various clinical conditions [19–21], thus providing the rationale for applications of split-belt walking to rehabilitation.

A considerable portion of the literature on split-belt treadmill walking has focused on post-stroke hemiparesis and changes in spatiotemporal gait parameters, with special reference to the associated asymmetry between subsequent steps [4,6]. Nevertheless, direct evidence of asymmetric muscular work can be provided by measurements of dynamic parameters only, and dynamic symmetry appears as a more relevant goal for rehabilitation. Spatiotemporal and kinematic symmetry may conceal dynamic asymmetries. Nearly perfect kinematic symmetry can be achieved by overloading the unimpaired lower limb in various clinical conditions, including post-stroke hemiparesis and hip arthritis [22,23], and following surgical procedures, such as above- and below-knee amputation [24], and knee rotationplasty [25]. In any case, kinematic asymmetries are driven by alterations of lower limb joint dynamics [26]: therefore, understanding these is fundamental to hypothesise the clinical applications of split-belt treadmills.

It is possible that the prevalent, purely kinematic approach taken in rehabilitation studies was inspired by the similarity in temporal asymmetries entailed by split-belt walking in healthy participants and ground walking in limping patients. In fact, in healthy participants, the limb on the faster treadmill belt "escapes" from ground contact like the patient's impaired limb does during ground walking. The necessary work and power are provided mainly by the rear plantar flexors, which are also the main drivers of walking overground [11]. As a rule, however, during ground walking, patients also show a longer anterior SL and lower power output on the affected "escaping" side [22,27], whereas healthy controls walking on split-belt treadmills show a shorter step and higher work and peak power on the "fast-escaping" side [7]. Thus, the therapeutic use of split-belt walking remains a controversial issue. Three major questions are still unanswered: does split-belt walking cause a long-lasting effect? Should the therapist aim at a spatiotemporal or at a dynamic symmetry? Should the impaired lower limb be placed on the faster or the slower belt?

The present study presents novel findings elucidating the relationships between temporal, spatial, and dynamic asymmetries during adaptation to split-belt walking and post-adaptation. The study aims at providing new insight on the rationale for the therapeutic use of this unusual form of gait.

2. Materials and Methods

2.1. Participants

Participant enrolment took place from March to October 2019 at the Laboratory of Research of Neuromotor Rehabilitation of the Istituto Auxologico Italiano in Milan (Italy). The inclusion criteria were the ability to wittingly sign the informed consent form, understand the instructions, complete the locomotor task, and being aged between 18 and 45 years. The exclusion criteria were any neurologic or orthopaedic condition affecting walking or balance, orthopaedic surgical procedures in the 18 months before the study, symptomatic spine conditions, pregnancy, and previous experience of walking on a split-belt treadmill. The participants were tested for their foot dominance using the revised Waterloo footedness questionnaire [28]. Twelve healthy adults (seven women), with a mean age of 27.2 (4.4) years, a mean height of 172 (4.4) cm, and a mean body weight of 66.7 (10.6) kg were recruited. One participant was left-footed. No side effects were recorded.

2.2. Instruments and Methods

The instruments used for recording and analysing joint kinematics and dynamics were described in a previous article [29]. Walking was analysed on a split-belt force-sensorised treadmill (model ADAL 3D; Médical Développement, Andrézieux-Bouthéon, France) embedded in the floor. It consisted of two parallel, independent half-treadmills, each mounted on four three-dimensional (3D) piezoelectric force sensors (KI 9048B; Kistler, Winterthur, Switzerland). The velocity can increase up to 2.7 m s^{-1} (i.e., 10 km h^{-1}) and be regulated in 0.1 m s^{-1} intervals. Changes in the belts' velocities take 2–3 s to develop fully. Force and belt velocity signals were sampled at 250 Hz. Joint kinematics were recorded through an optoelectronic method as per the Davis anthropometric model [30]. The 3D displacement of the markers was captured using eight near-infrared stroboscopic cameras (Smart-D optoelectronic system; BTS Bioengineering Spa, Milan, Italy; sampling rate 250 Hz). This procedure enabled the estimation of the ankle, knee, and hip excursions in 3D.

2.3. Experimental Protocol

The experimental protocol in this study was consistent with the one adopted by Reisman et al. [16] in their first published study on split-belt treadmill walking in healthy participants. The same design was implemented in most studies on both healthy and pathologic participants [4,15,31].

According to this protocol, participants must walk on a split-belt treadmill in either a "tied" or "split" condition, i.e., with the two belts moving at the same or different velocities, respectively. The split-belt walking test is preceded by tied-walking habituation, with the belts rotating at several distinct velocities. The habituation phase is usually followed by a few minutes of resting pause. In the subsequent testing session, usually after an initial brief tied-walking period (baseline phase), the velocity of one of the two belts is increased (or, rarely, decreased [32]) while the velocity of the other belt remains unchanged, resulting in a split-walking period (adaptation phase). In the following phase, tied-walking is restored (post-adaptation), as the velocity of the "fast" belt is decreased to the velocity of the slower belt. Several ratios between the two belts' velocities have been tested during the adaptation phase (usually 1:2, 1:3, or 1:4 ratios). Some studies included a resting pause of a few minutes between the adaptation and post-adaptation.

2.4. Experimental Session

Participants wore a t-shirt, shorts, and light gym shoes. The participants were then equipped with visual skin markers and electrodes (see below), and their height and weight were measured on a precision scale (precision: 2 mm and 0.05 kg, respectively). Afterwards, the walking test on the treadmill began (see below for further details). Participants were requested not to look at the treadmill belts during walking but to focus their vision on a black spot (8 cm in diameter) at about a 2 m distance and 1.65 m height on a white wall in front of the treadmill. Contrary to other studies [16,32–34], the participants had to walk freely with no external support (e.g., a suspension harness or handrails). A verbal warning was given to the participant before any change in the belts' velocities. Unlike other studies [16,33], no pauses were allowed between the baseline, adaptation, and post-adaptation phases. The participants' preparation and walking tests lasted, on average, about 60 min.

2.5. Tagging Walking Patterns and Lower Limbs

The velocity of each treadmill belt was given in metres per second and assigned a two-digit tag with no decimal dot (e.g., 04 denoting a velocity of 0.4 m s^{-1}). The combined velocities of the two belts were assigned a four-digit tag, the former pair of digits referring to the belt under the non-dominant lower limb. For instance, the tag 0412 referred to a walking modality during which one belt rotated at 0.4 m s^{-1} (non-dominant leg), and the other rotated at 1.2 m s^{-1}. Regardless of the belts' velocities, walking modalities were also referred to as "tied" when both treadmill belts rotated at the same velocity or "split" otherwise [7]. The return to the tied modality after split-walking was marked by a "post" suffix, e.g., 0404post. The habituation and baseline phases were identified by the prefixes h- and b-, respectively. The initial and final phases of the adaptation and post-adaptation (see below) were identified by the prefixes i- and f-, respectively. Of note, as "initial" here, the 7th to 12th strides after the transition between gait modalities (i.e., tied to split, split to tied) were considered. The mechanics of the very early adaptation were highly erratic across subjects and were not the target of the present study (although it is of interest that this transition phase still has not been the target of dedicated studies to the best of the authors' knowledge). For instance, i-0412 indicates the 7th to 12th strides within the adaptation phase, and f-0404post indicates the last six-stride set of the post-adaptation phase. In this study, during split tests, the dominant limb was always placed on the faster belt and defined, for simplicity, as the "fast" limb in opposition to the contralateral non-dominant "slow" limb.

2.6. Tagging the Test Phases

The test phases (in chronological order) were organised as follows:

(1) Habituation: 3 min tied-walking at a velocity increasing from 0.2 m s^{-1} (h-0202) to 1.2 m s^{-1} (h-1212). These changes occurred in 0.2 m s^{-1} increments, with each tested velocity lasting about 30 s. At the end of this phase, rest for about 1–2 min was allowed.
(2) Baseline: 30 s tied-walking at 0.4 m s^{-1} (b-0404).
(3) Adaptation: 15 min split-walking; at the end of the baseline, the belt's velocity under the dominant lower limb was increased to 1.2 m s^{-1} (0412).
(4) For analysis, the adaptation phase was further divided into two phases:
    (a) initial adaptation: including the 7th to 12th strides (i-0412).
    (b) final adaptation: including the last six strides (f-0412).
(5) Post-adaptation: Return to tied-walking at 0.4 m s^{-1} for 5 min (0404post).

For analysis, the post-adaptation phase was further divided into two phases:

(a) initial post-adaptation: the 7th to 12th strides (i-0404post).
(b) final post-adaptation: the last six strides (f-0404post).

2.7. Tagging Spatiotemporal Walking Variables

Here follows a glossary of the definitions adopted:

- Step: the ensemble of kinematic and dynamic events taking place between two subsequent foot–ground contacts.
- SL: the sagittal distance between the markers put on the lateral malleolus of the posterior and anterior feet at the ground strike of the anterior foot.
- Side of the step: the side of the posterior foot during a double stance.
- Single stance time (SST): for each lower limb, the time interval during which a vertical ground reaction $\geq$ 30 N was recorded under the limb.
- Double stance time (DST): the time interval during which a vertical ground reaction $\geq$ 30 N was recorded under both lower limbs.
- Side of the double stance time (pDST): the side of the posterior foot.

It must be highlighted that the side of the step and side of the DST were defined in relation to the posterior foot, at variance with the dominant convention. This alternative convention was adopted here to focus attention on the dynamics, rather than the kinematics, of walking [7]; during the double stance, the posterior foot provides most of the power needed to propel the body forward [35].

2.8. Correction of Drift in Force Signals

The treadmill used in the present study rested on eight 3D piezoelectric force transducers (four under each belt). A slow "positive" signal drift occurs with piezoelectric force transducers, mostly independent from the force load [36,37]. This may lead to an overestimation of forces with longer measurement durations. In the treadmill used in the present study, given the duration of the tests, such an error was detectable only for the vertical forces [29]. In a 20 min walking session, this drift may lead to an extra-vertical force of about 130 N. To offset the signal drift, a custom-made algorithm was developed and adopted in post-processing (tracings and algorithm available on request).

2.9. Computing the Velocity of the Body's Centre of Mass

Comparing tied and split-walking requires considering the actual velocity of the body system, as represented by its centre of mass (CoM). A common mistake in the literature is assuming that the CoM velocity equals the mean value between the two belts' velocities.

Instead, on each stride, the CoM velocity must be computed as the sum of the velocities of each belt, weighted by the percentage of stride time during which the resulting ground reaction force originated from each belt [38]. Therefore, the mean CoM velocity across six strides was computed for each subject during each phase of the split-walking test (i.e., i-0412 and f-0412).

2.10. Test Sequencing

Table 1 summarises the sequence adopted in the experimental session. The same velocities were imposed on all participants.

Table 1. The sequence of walking modalities that were adopted in the experimental session.

Habituation		Baseline	Adaptation	Post-Adaptation
h-0202 → h-0404 → h-0606 → h-0808 → h-1010 → h-1212	Rest	b-0404	0412	0404post
3 min	2 min	30 s	15 min	5 min

2.11. Data Analysis

Joint torque and power were computed through the spatiotemporal synchronisation of the ground reaction force vectors and the position of the model-estimated (see below) joint centres of rotation. Only the sagittal plane was considered in the analysis of the kinematic and dynamic variables.

Joint power was computed as the product of the joint torque and joint rotation speed [29]. Power was defined as "positive" or "generated" when the joint torque and rotation speed shared the same direction (i.e., when the agonist muscles were actively shortening) and as "negative" or "absorbed" otherwise. Positive work was computed as the integral of the generated (positive) power over time [27,39].

Strides from throughout the session were analysed and graphically represented. All signals were synchronised and analysed offline using algorithms that are available within the SMART Software Suite (BTS Bioengineering Spa, Milan, Italy). The stride time was normalised to 100 time points.

For the analysis, only the results from six subsequent strides in each phase were considered: six baseline strides, and six strides defining the initial and final phases of adaptation and post-adaptation, respectively (as described above). Data were averaged for each participant and then grand-averaged across all participants. For habituation, only the 0808 velocity pattern was taken into account.

2.12. Statistical Analysis

A sample size of 12 people was adopted in this study. This was considered to be sufficient for a reliable estimate of the parameters. The known and constant average velocity imposed by the treadmill guaranteed a very high reproducibility within participants across subsequent steps [7,27,29]. Continuous data are given as mean (standard deviation (SD)).

The Shapiro–Wilk test was used to assess the normality of the distribution of spatiotemporal and dynamic variables. Asymmetries between the right and the left steps were quantified as ratios (log-linearized) [7,23]. Ratios are extremely sensitive to outlier values. Therefore, among log-ratios only, outlier values, which were defined as values that exceeded the mean $\pm$ 2 SD, were excluded from the analysis. For the habituation and baseline values, symmetry was defined as the log-ratio showing 95% confidence limits encasing 0 (i.e., a ratio of 1). The main research question was whether, and by what amount, the symmetry status was violated or restored across the test conditions.

The fast/slow limb log-ratios in each test condition were also compared across conditions. A repeated analysis of variance (ANOVA) was applied to the fast/slow limb log-ratios of the following variables: ankle peak power (APP), ankle work (AW), posterior SL (pSL), SST, and pDST across five walking modalities (b-0404, i-0412, f-0412, i-0404post, f-0404post). In the case of a significant ANOVA model, Tukey's post hoc test was run on pairwise differences between modalities. The variance explanation was computed as the η^2 coefficient [40]. Significance was set at $p < 0.05$, and the Bonferroni correction for multiplicity was applied.

The changes in these parameters on either side over time were graphically described by encasing their moving average (see results, Figure 1) in "bands" with widths of ±1 SD. This allowed the analyst to perceive at a glance the overlap between the time courses of parameters of opposite sides. The underlying idea is that any overlap is equivalent to a non-significant difference at $p < 0.05$ (i.e., mean values sharing an interval of 2 SD) [29,39].

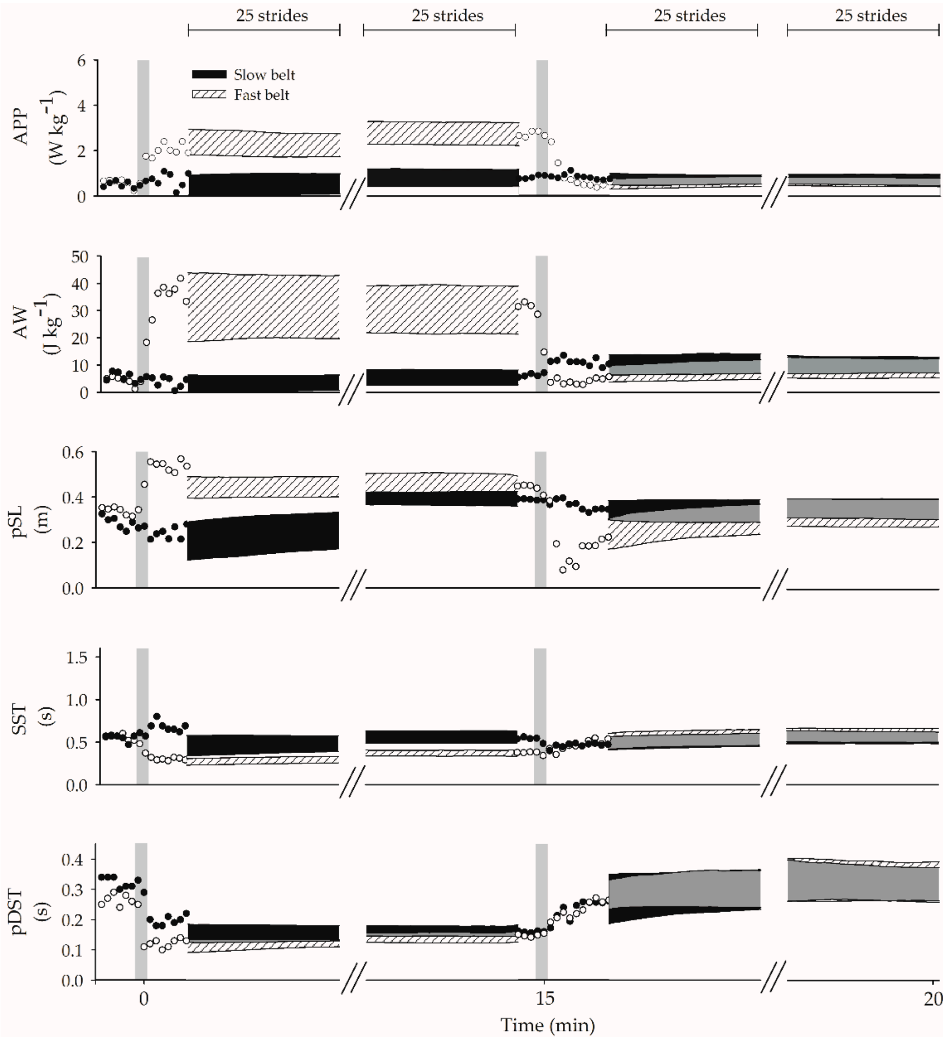

Figure 1. Dynamic and spatiotemporal parameters of the sample of 12 healthy participants during the baseline, adaptation, and post-adaptation phases. From top to bottom, for the ankle peak power (APP), ankle work (AW), posterior step length (pSL), single stance time (SST), and posterior double stance time (pDST), the bands encompassed the mean (±SD) values calculated across the 12 participants. A moving average was applied using a 30-stride window. For the sake of clarity, in the initial tied modality and during the transition between the adaptation and post-adaptation phases, the bands are substituted by dots signifying the grand-mean values. The black bands and dots refer to the slow limb (belt rotating at 0.4 m s^{-1}), while the dashed bands and the white dots refer to the fast limb (belt rotating at 0.4 m s^{-1} during the baseline and post-adaptation and 1.2 m s^{-1} during adaptation). The dark grey bands represent the overlaps between values recorded on either belt, thus highlighting the degree of the parameters' symmetry between sides. The light grey vertical bands highlight the strides occurring during the transitions between the belts' velocities, which took 2–3 s to fully develop. The time axis is interrupted (oblique segments) for visual clarity (note, on the abscissa, the different overall duration of the adaptation and post-adaptation phases). The overall number of steps during the same time window may differ across participants (not shown).

2.13. Computations

Statistical analysis and graphical representation were performed using MATLAB (version 2019b; Math Works Inc., Natick, MA, USA), STATA (version 14.0; STATA Corp., College Station, TA, USA), and SigmaPlot software (version 12.0; Systat Software Inc., San Jose, CA, USA).

2.14. Ethical Approval

The experiments were conducted according to the Declaration of Helsinki of the World Medical Association (2003). All participants provided written informed consent. The study was approved by the local institutional ethics committee.

3. Results

3.1. Dynamic and Spatiotemporal Observations

Figure 1 illustrates, from top to bottom, the temporal course of the dynamic parameters of the ankle joint's sagittal motion (APP, and AW) and the spatiotemporal parameters of walking (pSL, SST, and pDST) for the fast and slow limbs during the baseline, adaptation, and post-adaptation.

Numerical details of dynamic and spatiotemporal parameters for both lower limbs are provided in Table 2.

Table 2. Dynamic and spatiotemporal walking parameters of the sample of 12 healthy participants.

	APP (W kg^{-1})				AW (J kg^{-1})				pSL (m)				SST (s)				pDST (s)			
	Fast		Slow		Fast		Slow		Fast		Slow		Fast		Slow		Fast		Slow	
	Mean	SD	Mean	SD	Mean	SD	Mean	SD	Mean	SD	Mean	SD	Mean	SD	Mean	SD	Mean	SD	Mean	SD
h-0808	1.98	0.46	1.66	0.51	17.82	4.39	15.9	5.31	0.46	0.05	0.44	0.05	0.45	0.05	0.44	0.06	0.16	0.02	0.16	0.02
b-0404	0.75	0.26	0.69	0.28	9.61	5.09	8.61	4.62	0.33	0.06	0.31	0.06	0.53	0.07	0.53	0.08	0.31	0.07	0.31	0.07
i-0412	2.28	0.61	0.52	0.51	30.18	12.90	3.32	3.62	0.44	0.05	0.23	0.09	0.28	0.04	0.47	0.12	0.12	0.02	0.15	0.03
f-0412	2.65	0.72	0.84	0.40	28.81	9.32	5.85	3.21	0.44	0.06	0.39	0.05	0.37	0.07	0.56	0.07	0.15	0.03	0.16	0.03
i-0404post	0.49	0.38	0.70	0.35	6.04	5.84	10.19	5.29	0.35	0.04	0.22	0.09	0.51	0.13	0.46	0.09	0.26	0.12	0.27	0.06
f-0404post	0.60	0.21	0.66	0.25	8.97	3.97	9.51	4.12	0.34	0.06	0.34	0.05	0.59	0.10	0.55	0.10	0.32	0.07	0.33	0.08

From left to right, data refer to the ankle peak power (APP), ankle work (AW), posterior step length (pSL), single stance time (SST), and posterior double stance time (pDST) on the fast and slow treadmill belts. The following rows (second-to-last column from left) show the mean and standard deviation (SD) of six subsequent strides. Leftmost column: the first and second couples of numerals give the belts' velocities under the non-dominant and the dominant legs, respectively (with no decimal point: e.g., 0412 indicates 0.4 m s^{-1} and 1.2 m s^{-1}, respectively). Prefixes: h—habituation; b—baseline; i—7th to 12th strides; f—last six strides. Suffix: post—post-adaptation.

3.2. Graphic Description of Dynamic Changes

Figure 1 shows that for each parameter during the post-adaptation, the time course of values recorded on either side overlapped at least partially, indicating a non-significant difference. Nevertheless, different trends could be detected across the parameters. Given that standard deviations and p-values are highly dependent on the sample size, all graphically visible differences between sides are described, regardless of their significance.

3.2.1. APP

Figure 1 (first row) shows that the APP underwent a brisk increase on the fast side from the very beginning of the adaptation phase. At the end of the adaptation phase, the mean APP values mildly increased on both sides. At the very beginning of the post-adaptation phase, the APP returned to the baseline values in a matter of 5–6 strides. Data variability (see the "thickness" of the horizontal bands) decreased on both sides. A mild trend towards lowered APP could be detected on the formerly fast side compared to the formerly slow one (see the thin black and dashed "deburring" above and below the grey band, respectively). This subtle after-effect faded within 5 min of the post-adaptation.

3.2.2. AW

During the whole adaptation phase, the AW (second row) increased remarkably on the fast side and only mildly on the slow side. During the post-adaptation, baseline values were recovered in a few strides. Data variability remained low for the formerly slow side and decreased for the formerly fast side. A mild tendency for a reversal of the asymmetry was observed, fading during the 5 min post-adaptation (see again the partial overlap between the dashed and black bands).

3.3. Graphic Description of Spatiotemporal Changes

3.3.1. pSL

Figure 1 (third row) shows that the pSL in the fast limb rapidly increased at the beginning of the adaptation phase. This phenomenon can be interpreted as the effect of a "dragging" action by the fast belt on the corresponding lower limb than by the slow belt. In contrast, on the slow side, the pSL showed an initial reduction and then gradually increased across the duration of the adaptation phase, approaching the contralateral values (of note, the 1 SD limits do not graphically overlap). This implies that the stride length increased. During the post-adaptation, baseline values were soon recovered. For a couple of minutes, however, the pSL became markedly lower on the formerly fast side, as evidenced by the small overlap between the superior black and the inferior dashed bands (a clear after-effect). Symmetry was reached at the end of the post-adaptation (see below for the results of the statistical analysis).

3.3.2. SST

At the beginning of the adaptation phase, the SST of the fast limb (Figure 1, fourth row) suddenly decreased and then mildly increased with time. On the slower side, the SST showed a sudden fleeting increase, returned to baseline values, and then showed a moderate increase with time. During the whole adaptation phase, the SST remained lower on the faster side, matching the visual impression of "escape" claudication. During the post-adaptation, the baseline values seemed to be reached within a few strides.

3.3.3. pDST

During the initial adaptation phase, the pDST decreased on both sides and then increased moderately with time (Figure 1, fifth row). A tendency towards partial asymmetry (the pDST was shorter on the fast side) was observed. During the post-adaptation, the baseline values were recovered in approximately 10 strides. Of note, data variability was much higher for both sides during the post-adaptation than during the adaptation.

Numeric summaries of the changes in parameters across the walking phases are given in Table 3.

Table 3. Repeated ANOVA model, partial variance explanation (η^2), and Tukey's post hoc test on comparisons between pairs between the fast/slow limb ratios (linearised through a log transformation) of the ankle peak power (APP), ankle work (AW), posterior step length (pSL), single stance time (SST), and posterior double stance time (pDST) in the five test modalities.

	APP (*n* = 346; *n*-Out = 14)	**AW (*n* = 349; *n*-Out = 11)**	**pSL (*n* = 344; *n*-Out = 16)**	**SST (*n* = 343; *n*-Out = 17)**	**pDST (*n* = 344; *n*-Out = 16)**
Baseline phase values, mean (95% C.I.)	0.07 (−0.22 ÷ 0.33)	0.06 (−0.27 ÷ 0.35)	0.03 (−0.05 ÷ 0.11)	0.01 (−0.07 ÷ 0.07)	0.00 (−0.10 ÷ 0.10)
Repeated ANOVA model					
R^2	0.89	0.91	0.94	0.90	0.76
Bonferroni-corrected *p*-value = 0.01					
Model	0.00 *	0.00 *	0.00 *	0.00 *	0.00 *
Test modality	0.00 *	0.00 *	0.00 *	0.00 *	0.00 *
η^2					
Model	0.89	0.91	0.94	0.90	0.76
Test modality	0.85	0.89	0.92	0.87	0.50

Table 3. *Cont.*

	APP (n = 346; n-Out = 14)	AW (n = 349; n-Out = 11)	pSL (n = 344; n-Out = 16)	SST (n = 343; n-Out = 17)	pDST (n = 344; n-Out = 16)
Tukey's post hoc test					
b-0404 vs. i-0412	**0.00** $^{\#}$	**0.00** $^{\#}$	**0.00** $^{\#}$	**0.00** $^{\#}$	**0.00** $^{\#}$
b-0404 vs. f-0412	0.00 $^{\#}$	0.00 $^{\#}$	0.77	0.00 $^{\#}$	0.13
b-0404 vs. i-0404post	0.00 $^{\#}$	0.00 $^{\#}$	0.00 $^{\#}$	0.02 $^{\#}$	0.00 $^{\#}$
b-0404 vs. f-0404post	**0.25**	**0.45**	**0.33**	**0.71**	**0.99**
i-0412 vs. f-0412	0.00 $^{\#}$	0.00 $^{\#}$	0.00 $^{\#}$	0.02 $^{\#}$	0.00 $^{\#}$
i-0412 vs. i-0404post	0.00 $^{\#}$	0.00 $^{\#}$	0.00 $^{\#}$	0.00 $^{\#}$	0.00 $^{\#}$
i-0412 vs. f-0404post	0.00 $^{\#}$	0.00 $^{\#}$	0.00 $^{\#}$	0.00 $^{\#}$	0.00 $^{\#}$
f-0412 vs. i-0404post	0.00 $^{\#}$	0.00 $^{\#}$	0.00 $^{\#}$	0.00 $^{\#}$	0.00 $^{\#}$
f-0412 vs. f-0404post	0.00 $^{\#}$	0.00 $^{\#}$	0.02 $^{\#}$	0.00 $^{\#}$	0.35
i-0404post vs. f-0404post	0.03 $^{\#}$	0.00 $^{\#}$	0.00 $^{\#}$	0.31	0.00 $^{\#}$

Labelling of variables (upper row) and contrasted data sets (leftmost column, 10 lowest rows), as in Table 1. Outliers, defined as values that exceeded mean ± 2 standard deviations, were excluded from the analysis. n: number of observations without outliers. n-Out: number of outliers. $^{\#}$: significant at $p < 0.05$. *: significant at $p < 0.01$ after a Bonferroni correction for multiplicity. Bold characters highlight the contrasts between the basal and initial split conditions and between the basal and final post-adaptation conditions, respectively.

After the Tukey's post hoc test analysis, it may be seen that for all dynamic and kinematic parameters, the contrasts for fast/slow limb ratios were significant between the basal condition (b-0404) and initial adaptation (i-0412). In contrast, all parameters were non-significant between the baseline and final post-adaptation (b-0404 and f-0404post, respectively). For clarity, data in the rows referring to these contrasts are presented with bold characters in Table 3. Therefore, the mild tendency for reversal of the asymmetry observed for the five parameters in Figure 1 (see the incomplete overlap between the dashed and black bands) was very short-lasting. The numeric statistical analysis could not disconfirm the null hypothesis of the restoration of symmetry (i.e., the absence of any after-effect must be assumed) by the end of the post-adaptation phase (see Section 4).

3.4. Graphic Summary of the Time Course of Walking Parameters

Figure 2 further summarises the time course of the walking parameters across the walking phases and simplifies the information provided in Figure 1 (point values are given, and the initial post-adaptation is omitted); in addition, the panels give complementary information: (a) the means of parameters during tied-walking at 0.8 m s^{-1} (h-0808) (the first column of symbols from left, see also Table 1) and (b) the actual velocity of the body system, as represented by its CoM, during initial and final adaptation (see Section 2.9). This velocity is given at the top of the fourth and fifth columns of symbols from the left (as a median across subjects: 0.72 m s^{-1} and 0.67 m s^{-1} in i-0412 and f-0412, respectively).

During the initial adaptation, the median walking velocity (i.e., the velocity of the body's CoM) was lower than the intermediate velocity between the two belts (0.72 vs. 0.80 m s^{-1}, respectively), and it became even lower (0.67 m s^{-1}) at the end of this phase. Boldly stated, on average, the participants preferred propelling the body system from the slower belt rather than from the faster belt. This may explain why some kinematic and temporal parameters were slightly lower during the initial adaptation (pSL, SST, pDST) and final adaptation (pSL), compared to tied-walking at 0.8 m s^{-1}. This difference did not seem to be large enough to cause major differences in dynamic parameters. Figure 2 makes it evident that no substantial changes occurred between the final post-adaptation and baseline, either in the parameters' average values or in their variability.

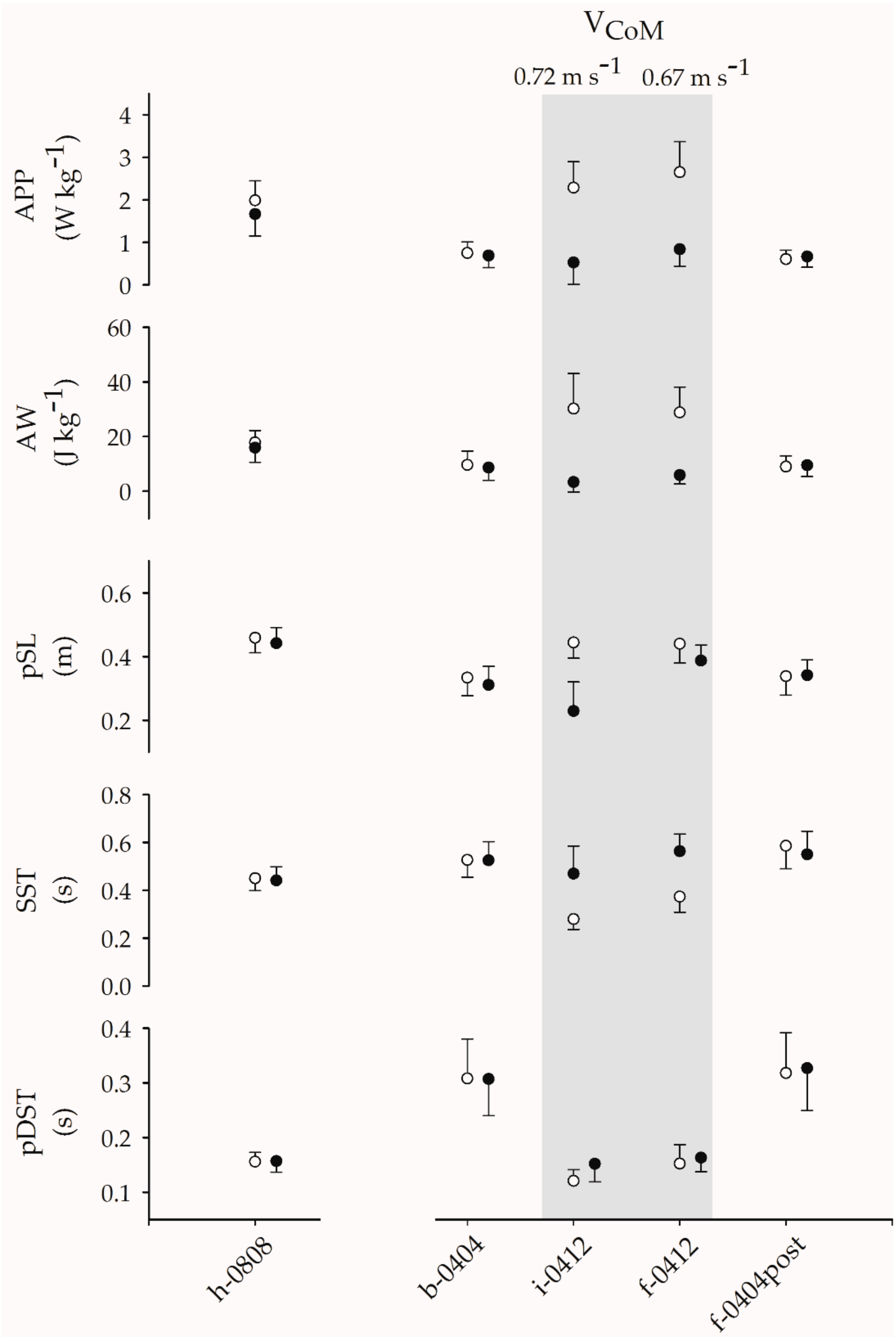

Figure 2. The ordinate shows the walking parameters (as in Figure 1 and Table 2): from top to bottom, ankle peak power (APP), ankle work (AW), posterior step length (pSL), single stance time (SST), and posterior double stance time (pDST). Symbols represent the results that were computed as grand averages (standard deviation (SD)) across 6 consecutive strides for each participant and across 12 participants (n = 72). White and black dots refer to the fast and the slow lower limbs, respectively. The columns (see abscissa) from left to right refer to the walking phases (see the legend in Figure 1). The vertical grey band encases the adaptation phase. Actual median velocities of the centre of mass (V_{CoM}) in the initial and final adaptation phases are given above the grey band.

Both the APP and AW showed a greater variance during adaptation than post-adaptation, which might reflect the differences in the actual velocity of the CoM across subjects [38]. Among the spatiotemporal parameters, the pDST, contrary to the dynamic parameters, showed a greater variance on both sides during the post-adaptation than during the adaptation. This might reflect the shortening of the SST during the adaptation, limiting the variance in absolute terms.

4. Discussion

The present results seem to inspire some suggestions concerning the physiology and clinical application of this different form of locomotion.

It is known that unilateral impairments of both neurologic and orthopaedic origins (e.g., hemiparesis after stroke, or above- or below-knee lower limb amputation) conceal severe dynamic asymmetries, even when they present with near-symmetrical kinematics. The unaffected lower limb may provide extra propulsive power, compensating for the power deficit from the affected lower limb (whatever its origin). The overall energy expenditure and work production allowing for the transfer of the body's CoM may remain unaffected [35,41]. In the long run, however, power asymmetry leads to a noxious overload of the unaffected limb and atrophy of the spared, impaired limb [35]. Paradoxically, in many instances, the impaired lower limb is still able to provide additional power in more demanding conditions (e.g., while walking faster or uphill [42,43], during crouch gait [27], or when placed on the faster belt of a split-belt treadmill [7]), yet it always provides less muscle power than the unaffected one. Power asymmetry, therefore, seems to be the core of these impairments rather than weakness itself.

As a rule, during pathologic claudication of any origin, the impaired lower limb shows a longer "anterior" step and a shorter stance time than the unaffected lower limb ("escape" limp). This behaviour is consistent with the impaired limb's lower contribution of power to the propulsion of the body's CoM from behind [7,27]. However, the analogy with the asymmetric gait induced by split-belt walking in healthy subjects is partly misleading. In healthy subjects on split-belt treadmills, the fast limb indeed shows a shorter stance time (resembling the aforementioned "escape" limp) but, at the same time, its posterior SL is increased as it is dragged backwards by the faster belt relative to the slower belt (see Section 1). Notwithstanding this peculiarity, this artificial form of locomotion has been firmly proposed as a treatment for "symmetrising" SL in pathologic walking [3,6,18,19,44]. Most authors agreed that the limb showing the longest anterior SL (as a rule, the impaired lower limb) should be placed on the slower belt, thus enhancing the pathologic asymmetry of SL during the split-walking session. Symmetry is expected as an after-effect [4–6]. Actually, the evidence for the effectiveness of this paradigm, even when applied to repeated sessions, remains scarce and elusive. Improved spatial symmetry alone, whenever attained, does not appear to result in substantial functional improvements for post-stroke patients (e.g., in balance, number of steps per day, participation, and community mobility) [45]. Evidence from repeated split-belt treadmill treatments is still very limited [3,44], and it remains to be determined whether persistent results can be obtained after repeated sessions. To the best of our knowledge, only one study has reported persisting results at 3 months after 4 weeks of split-belt treadmill training with three sessions of treatment per week [3]. Criticisms may thus be raised about the paradigm itself: (a) it reflects the clinical relevance assigned to SL asymmetry and (b) it implies "betting" on the therapeutic role of the after-effect, which is characterised by a reversal, or at least a decrease, of the pathologic SL asymmetry. Another clinical rationale might be provided by the error-augmentation principle: once a motor correction is triggered, it can trespass the baseline condition and point towards normality [19,46]. Whichever the rationale, the opposite effect of split-walking on stance time is neglected. After post-adaptation, the impaired limb should be driven to provide a smaller anterior step (the desired effect) but also an even shorter stance time (thus enhancing the claudication). Further study is required to elucidate the optimal target of this "symmetrizing" approach.

Split-belt walking provides a doubtful therapy for dynamic asymmetry as well. Adaptation to split-walking implies dynamic asymmetries of much greater magnitude than spatiotemporal asymmetries (Figures 1 and 2). This notwithstanding, when the tied-belt condition was restored, the after-effect on all of the formerly asymmetric variables, either dynamic or spatiotemporal, was very small and, in any case, extremely short-lived (this is consistent with previous studies on healthy participants [16,32]).

The popular idea [16,47,48] of an analogy with the also short-lasting, well-known after-effect of visuomotor tasks should be questioned. The typical experimental paradigms were focused on pointing tasks during and after the application of prismatic lenses displacing the visual field [49–52]. Both split-walking and prism adaptation seem to promote primarily implicit motor learning processes [53,54]. However, differences between the walking and pointing tasks should be highlighted; in pointing experiments, the task consists of the precise displacement of a discrete body segment (one upper limb); balance and energetic constraints are of minor relevance if any. By contrast, the split-belt walking task involves displacing the body system as a whole. Most lower limb movements are not consciously controlled but are arranged in multisegmental patterns [55] that are steered by subcortical sensory-motor circuits [56]. In addition, two homologous body segments (the lower limbs) must cooperate and the unimpaired one can compensate for unilateral impairments. Balance and energetic constraints are factors of the utmost relevance, where both must enter the equation for optimising the overall performance. This implies that in pointing tasks, the after-effect would reflect the short-lived retention of the processes aimed at error correction, while in split-belt walking, the after-effect might reflect the short time delay required for the automatic, reflex-driven reset of an unconscious, multifactorial postural adaptation.

Curiously enough, the current literature has neglected the potential therapeutic use of split-walking as a classical form of potentiation of the impaired lower limb. Ankle moment and the foot's position relative to the CoM at terminal stance contribute to the lower limb's propulsive force [57], and the split-belt paradigm provides an instrument for experimentally manipulating both the dynamics and kinematics of the ankle joint. In this view, according to a forced-use paradigm [27], the impaired side should be placed on the faster belt to improve the power and work provided by the affected lower limb and increase its pSL during the adaptation phase, thus aiming at correcting the dynamic asymmetry. This change in perspective would imply "betting" on the effect rather than on the after-effect of the treatment. Basic training principles could be easily respected, including task specificity, overload, progression in difficulty, individualisation, and task variability [58].

The analysis of the velocity of the CoM during split-walking may provide new insights on previous evidence. An earlier study has shown a reduction in net metabolic power and lower limbs' muscle surface electromyographic activity during the adaptation phase of split-walking [32]. Then, other studies described the final adaptation phase as economical compared to the initial adaptation phase by taking into account the mechanical work performed by the lower limbs [14] and the energy expenditure estimated from the rate of oxygen consumption [59]. This energetic optimisation throughout the adaptation phase has been interpreted as being associated with the gradual symmetrisation in SL [32]. Nonetheless, SL symmetrisation is barely the core of the adaptation phenomenon, although kinematic changes likely represent its most visible by-product. Instead, during split-belt walking, the attention should be shifted to the way participants can modify the forces exerted against each belt across subsequent strides, as can be recorded by force-sensorised split-belt treadmills. The belts' velocities, in themselves invariant, may be associated with individual changes in CoM velocity from stride to stride throughout the split-walking trial [38]. A reduction in the median CoM velocity, as described in the present study, might eventually lead to more economical walking. Further studies on the energetic parameters of the CoM during split-belt treadmill walking are needed.

The limitations of the study cannot be overemphasised. First, only the dynamic parameters of the ankle joint were taken into account, and those of the knee and hip were neglected. On the other hand, hip flexor muscles were shown to generate much lower work and power than the ankle plantar flexors during both tied- and split-walking at velocities ranging from 0.4 to 1.2 m s^{-1} [7]. Second, the small sample size used in this study might be a source of concern. On the other hand, in previous studies, this sample size was shown to be sufficient to highlight significant differences in both spatiotemporal and dynamic parameters, thanks to the low variance allowed by the imposed and constant velocities of the treadmill belts (see Section 2.12). In addition, to be compared with normative data, future cohorts of patients would be unlikely to be numerous given the restrictive inclusion criteria and the complexity of the experimental paradigm. Third, only one combination of velocities was tested in the split condition (i.e., 0.4 and 1.2 m s^{-1}, a 1:3 ratio). This ratio was consistent with published paradigms, which have adopted velocity ratios ranging from 1:1.2 to 1:4. It remains true, however, that higher ratios might trigger different behaviours. Finally, the dynamic response in cases of impairments might not necessarily follow the one observed here in healthy participants; therefore, direct experimentation on impaired patients cannot be overlooked.

Our results suggest some topics for future research. First, our results confirmed [38] that force-sensorised treadmills should always be adopted during split-belt walking studies to assess the dynamic parameters (not only the spatiotemporal ones) of the lower limb joints during adaptation and post-adaptation. Moreover, force sensors under each belt are needed to estimate the actual velocity of the CoM. Second, the actual velocity of the CoM during split-walking should be considered, both at a group level and in individual participants, to run comparisons. To date, comparisons were only performed with tied-walking at the velocity of the slower or faster belt [14,16,26,32,60] or at the intermediate velocity between the two belts [7,14,32,48,59,61,62]. Third, trials should be performed by comparing the results obtained by assigning the impaired lower limb to the fast belt, not only to the slow belt. In either case, it has to be decided whether the study is aimed at investigating dynamic or spatial symmetry (see our ongoing trial, ClinicalTrials.gov, accessed on 1 May 2021 ID: NCT04635436). Fourth, the motion of the body's CoM should be analysed to interpret effects at the level of the lower joints in light of the changes induced at the whole-system level [35].

5. Conclusions

Split-belt walking in healthy adults mimics pathologic claudication in its temporal step asymmetries (shorter step time on the faster belt), but it provides the opposite spatial (step length) and dynamic (ankle joint power and work) asymmetries. Transient reversal of all of these asymmetries can be obtained as a very short-lasting after-effect during post-adaptation. The literature on hemiparetic gaits overlooks dynamic events and privileges the achievement of step length symmetry as an after-effect (impaired lower limb on the slower belt during split-belt walking) at the expense of a higher stance time asymmetry. Dynamic symmetry looks like a more fundamental rehabilitation goal. This goal might be achieved as a direct effect of split-belt walking (with the impaired lower limb on the faster belt). This approach would implement the overload and forced-use rehabilitation principles rather than more cognitive learning principles.

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Informed Consent Statement: Informed consent was obtained from all subjects involved in the study.

Data Availability Statement: Data are available upon request.

Conflicts of Interest: The authors declare no conflict of interest.

References

1. Dietz, V.; Zijlstra, W.; Duysens, J. Human neuronal interlimb coordination during split-belt locomotion. *Exp. Brain Res.* **1994**, *101*, 513–520. [CrossRef]
2. Reisman, D.S.; McLean, H.; Bastian, A.J. Split-belt treadmill training poststroke: A case study. *J. Neurol. Phys. Ther.* **2010**, *34*, 202–207. [CrossRef] [PubMed]
3. Reisman, D.S.; McLean, H.; Keller, J.; Danks, K.A.; Bastian, A.J. Repeated split-belt treadmill training improves poststroke step length asymmetry. *Neurorehabil. Neural Repair* **2013**, *27*, 460–468. [CrossRef]
4. Reisman, D.S.; Wityk, R.; Silver, K.; Bastian, A.J. Split-belt treadmill adaptation transfers to overground walking in persons poststroke. *Neurorehabil. Neural Repair* **2009**, *23*, 735–744. [CrossRef]
5. Kline, P.; Murray, A.; Mille, M.; Fields, T.; Christiansen, C. Error-augmentation gait training to improve gait symmetry in patients with non-traumatic lower limb amputation: A proof of concept study. *Prosthet. Orthot. Int.* **2019**, *43*, 426–433. [CrossRef] [PubMed]
6. Lauzière, S.; Mièville, C.; Betschart, M.; Duclos, C.; Aissaoui, R.; Nadeau, S. A more symmetrical gait after split-belt treadmill walking increases the effort in paretic plantar flexors in people post-stroke. *J. Rehabil. Med.* **2016**, *48*, 576–582. [CrossRef] [PubMed]
7. Tesio, L.; Malloggi, C.; Malfitano, C.; Coccetta, C.A.; Catino, L.; Rota, V. Limping on split-belt treadmills implies opposite kinematic and dynamic lower limb asymmetries. *Int. J. Rehabil. Res.* **2018**, *41*, 304–315. [CrossRef] [PubMed]
8. Sutherland, D.H. An electromyographic study of the plantar flexors of the ankle in normal walking on the level. *J. Bone Jt. Surg.* **1966**, *48*, 66–71. [CrossRef]
9. Meinders, M.; Gitter, A.; Czerniecki, J.M. The role of ankle plantar flexor muscle work during walking. *Scand. J. Rehabil. Med.* **1998**, *30*, 39–46. [CrossRef]
10. Kepple, T.M.; Siegel, K.L.; Stanhope, S.J. Relative contributions of the lower extremity joint moments to forward progression and support during gait. *Gait Posture* **1997**, *6*, 1–8. [CrossRef]
11. Zelik, K.E.; Adamczyk, P.G. A unified perspective on ankle push-off in human walking. *J. Exp. Biol.* **2016**, *219*, 3676–3683. [CrossRef] [PubMed]
12. Vervoort, D.; Rob Den Otter, A.; Buurke, T.J.W.; Vuillerme, N.; Hortobágyi, T.; Lamoth, C.J.C. Effects of aging and task prioritization on split-belt gait adaptation. *Front. Aging Neurosci.* **2019**, *11*, 1–12. [CrossRef] [PubMed]
13. Hinton, D.C.; Conradsson, D.; Bouyer, L.; Paquette, C. Does dual task placement and duration affect split-belt treadmill adaptation? *Gait Posture* **2020**, *75*, 115–120. [CrossRef] [PubMed]
14. Selgrade, B.P.; Thajchayapong, M.; Lee, G.E.; Toney, M.E.; Chang, Y.H. Changes in mechanical work during neural adaptation to asymmetric locomotion. *J. Exp. Biol.* **2017**, *220*, 2993–3000. [CrossRef]
15. Reisman, D.S.; Wityk, R.; Silver, K.; Bastian, A.J. Locomotor adaptation on a split-belt treadmill can improve walking symmetry post-stroke. *Brain* **2007**, *130*, 1861–1872. [CrossRef]
16. Reisman, D.S.; Block, H.J.; Bastian, A.J. Interlimb coordination during locomotion: What can be adapted and stored? *J. Neurophysiol.* **2005**, *94*, 2403–2415. [CrossRef]
17. Martin, T.A.; Keating, J.G.; Goodkin, H.P.; Bastian, A.J.; Thach, W.T. Throwing while looking through prisms II. Specificity and storage of multiple gaze-throw calibrations. *Brain* **1996**, *119*, 1199–1211. [CrossRef]
18. Kline, P.W.; Murray, A.M.; Miller, M.J.; So, N.; Fields, T.; Christiansen, C.L. Step length symmetry adaptation to split-belt treadmill walking after acquired non-traumatic transtibial amputation. *Gait Posture* **2020**, *80*, 162–167. [CrossRef]
19. Helm, E.E.; Reisman, D.S. The split-belt walking paradigm: Exploring motor learning and spatiotemporal asymmetry poststroke. *Phys. Med. Rehabil. Clin. N. Am.* **2015**, *26*, 703–713. [CrossRef]
20. Patterson, K.K.; Gage, W.H.; Brooks, D.; Black, S.E.; McIlroy, W.E. Evaluation of gait symmetry after stroke: A comparison of current methods and recommendations for standardization. *Gait Posture* **2010**, *31*, 241–246. [CrossRef] [PubMed]
21. Rudroff, T.; Proessl, F. Effects of muscle function and limb loading asymmetries on gait and balance in people with multiple sclerosis. *Front. Physiol.* **2018**, *15*, 531. [CrossRef]
22. Tesio, L.; Civaschi, P.; Tessari, L. Motion of the center of gravity of the body in clinical evaluation of gait. *Am. J. Phys. Med.* **1985**, *64*, 57–70.
23. Cavagna, G.A.; Tesio, L.; Fuchimoto, T.; Heglund, N.C. Ergometric evaluation of pathological gait. *J. Appl. Physiol.* **1983**, *55*, 607–613. [CrossRef] [PubMed]
24. Tesio, L.; Lanzi, D.; Detrembleur, C. The 3-D motion of the centre of gravity of the human body during level walking. II. Lower limb amputees. *Clin. Biomech.* **1998**, *13*, 83–90. [CrossRef]
25. Rota, V.; Benedetti, M.G.; Okita, Y.; Manfrini, M.; Tesio, L. Knee rotationplasty: Motion of the body centre of mass during walking. *Int. J. Rehabil. Res.* **2016**, *39*, 346–353. [CrossRef] [PubMed]
26. Roemmich, R.T.; Stegemöller, E.L.; Hass, C.J. Lower extremity sagittal joint moment production during split-belt treadmill walking. *J. Biomech.* **2012**, *45*, 2817–2821. [CrossRef]
27. Tesio, L.; Rota, V.; Malloggi, C.; Brugliera, L.; Catino, L. Crouch gait can be an effective form of forced-use/no constraint exercise for the paretic lower limb in stroke. *Int. J. Rehabil. Res.* **2017**, *40*, 254–267. [CrossRef] [PubMed]
28. Elias, L.J.; Bryden, M.P.; Bulman-Fleming, M.B. Footedness is a better predictor than is handedness of emotional lateralization. *Neuropsychologia* **1998**, *36*, 37–43. [CrossRef]

29. Tesio, L.; Rota, V. Gait analysis on split-belt force treadmills: Validation of an instrument. *Am. J. Phys. Med. Rehabil.* **2008**, *87*, 515–526. [CrossRef]
30. Davis, R.B.; Õunpuu, S.; Tyburski, D.; Gage, J.R. A gait analysis data collection and reduction technique. *Hum. Mov. Sci.* **1991**, *10*, 575–587. [CrossRef]
31. Ogawa, T.; Kawashima, N.; Ogata, T.; Nakazawa, K. Predictive control of ankle stiffness at heel contact is a key element of locomotor adaptation during split-belt treadmill walking in humans. *J. Neurophysiol.* **2014**, *111*, 722–732. [CrossRef]
32. Finley, J.M.; Bastian, A.J.; Gottschall, J.S. Learning to be economical: The energy cost of walking tracks motor adaptation. *J. Physiol.* **2013**, *591*, 1081–1095. [CrossRef]
33. Lauzière, S.; Miéville, C.; Betschart, M.; Duclos, C.; Aissaoui, R.; Nadeau, S. Plantarflexion moment is a contributor to step length after-effect following walking on a split-belt treadmill in individuals with stroke and healthy individuals. *J. Rehabil. Med.* **2014**, *46*, 849–857. [CrossRef] [PubMed]
34. Roper, J.A.; Roemmich, R.T.; Tillman, M.D.; Terza, M.J.; Hass, C.J. Split-belt treadmill walking alters lower extremity frontal plane mechanics. *J. Appl. Biomech.* **2017**, *33*, 256–260. [CrossRef] [PubMed]
35. Tesio, L.; Rota, V. The motion of the body center of mass during walking: A review oriented to clinical applications. *Front. Neurol.* **2019**, *10*. [CrossRef] [PubMed]
36. Bocian, M.; Brownjohn, J.M.W.; Racic, V.; Hester, D.; Quattrone, A.; Monnickendam, R. A framework for experimental determination of localised vertical pedestrian forces on full-scale structures using wireless attitude and heading reference systems. *J. Sound Vib.* **2016**, *376*, 217–243. [CrossRef]
37. Hurkmans, H.L.P.; Bussmann, J.B.J.; Benda, E.; Verhaar, J.A.N.; Stam, H.J. Accuracy and repeatability of the pedar mobile system in long-term vertical force measurements. *Gait Posture* **2006**, *23*, 118–125. [CrossRef]
38. Tesio, L.; Scarano, S.; Cerina, V.; Malloggi, C.; Catino, L. Velocity of the body center of mass during walking on split-belt treadmill. *Am. J. Phys. Med. Rehabil.* **2021**, *100*, 620–624. [CrossRef]
39. Tesio, L.; Malloggi, C.; Portinaro, N.M.; Catino, L.; Lovecchio, N.; Rota, V. Gait analysis on force treadmill in children: Comparison with results from ground-based force platforms. *Int. J. Rehabil. Res.* **2017**, *40*, 315–324. [CrossRef]
40. Cohen, J. Eta-squared and partial eta-squared in fixed factor Anova designs. *Educ. Psychol. Meas.* **1973**, *33*, 107–112. [CrossRef]
41. Tesio, L.; Roi, G.S.; Möller, F. Pathological gaits: Inefficiency is not a rule. *Clin. Biomech.* **1991**, *6*, 47–50. [CrossRef]
42. Milot, M.-H.; Nadeau, S.; Gravel, D. Muscular utilization of the plantarflexors, hip flexors and extensors in persons with hemiparesis walking at self-selected and maximal speeds. *J. Electromyogr. Kinesiol.* **2007**, *17*, 184–193. [CrossRef]
43. Werner, C.; Lindquist, A.R.; Bardeleben, A.; Hesse, S. The influence of treadmill inclination on the gait of ambulatory hemiparetic subjects. *Neurorehabil. Neural Repair* **2007**, *21*, 76–80. [CrossRef]
44. Betschart, M.; McFayden, B.J.; Nadeau, S. Lower limb joint moments on the fast belt contribute to a reduction of step length asymmetry over ground after split-belt treadmill training in stroke: A pilot study. *Physiother. Theory Pract.* **2020**, *36*, 989–999. [CrossRef]
45. Ryan, H.P.; Husted, C.; Lewek, M.D. Improving spatiotemporal gait asymmetry has limited functional benefit for individuals poststroke. *J. Neurol. Phys. Ther.* **2020**, *44*, 197–204. [CrossRef]
46. Patton, J.L.; Stoykov, M.E.; Kovic, M.; Mussa-Ivaldi, F.A. Evaluation of robotic training forces that either enhance or reduce error in chronic hemiparetic stroke survivors. *Exp. Brain Res.* **2006**, *168*, 368–383. [CrossRef]
47. Torres-Oviedo, G.; Bastian, A.J. Natural error patterns enable transfer of motor learning to novel contexts. *J. Neurophysiol.* **2012**, *107*, 346–356. [CrossRef] [PubMed]
48. Sánchez, N.; Park, S.; Finley, J.M. Evidence of energetic optimization during adaptation differs for metabolic, mechanical, and perceptual estimates of energetic cost. *Sci. Rep.* **2017**, *7*, 7682. [CrossRef]
49. Redding, G.M.; Wallace, B. Adaptive eye-hand coordination: Implications of prism adaptation for perceptual-motor organization. *Adv. Psychol.* **1992**, *85*, 105–127.
50. Rossetti, Y.; Koga, K.; Mano, T. Prismatic displacement of vision induces transient changes in the timing of eye-hand coordination. *Percept. Psychophys.* **1993**, *54*, 355–364. [CrossRef] [PubMed]
51. Rossetti, Y.; Rode, G.; Pisella, L.; Farné, A.; Li, L.; Boisson, D.; Perenin, R.S. Prism adaptation to a rightward optical deviation rehabilitates left hemispatial neglect. *Lett. Nat.* **1998**, *395*, 8–11. [CrossRef]
52. Redding, G.M.; Rossetti, Y.; Wallace, B. Applications of prism adaptation: A tutorial in theory and method. *Neurosci. Biobehav. Rev.* **2005**, *29*, 431–444. [CrossRef]
53. Kleynen, M.; Braun, S.M.; Bleijlevens, M.H.; Lexis, M.A.; Rasquin, S.M.; Halfens, J.; Wilson, M.R.; Beurskens, A.J.; Masters, R.S.W. Using a delphi technique to seek consensus regarding definitions, descriptions and classification of terms related to implicit and explicit forms of motor learning. *PLoS ONE* **2014**, *9*, e100227. [CrossRef]
54. French, M.A.; Morton, S.M.; Reisman, D.S. Use of explicit processes during a visually guided locomotor learning task predicts 24-h retention after stroke. *J. Neurophysiol.* **2021**, *125*, 211–222. [CrossRef]
55. Lacquaniti, F.; Ivanenko, Y.P.; Zago, M. Patterned control of human locomotion. *J. Physiol.* **2012**, *590*, 2189–2199. [CrossRef]
56. Grillner, S.; El Manira, A. Current principles of motor control, with special reference to vertebrate locomotion. *Physiol. Rev.* **2020**, *100*, 271–320. [CrossRef] [PubMed]
57. Hsiao, H.Y.; Knarr, B.A.; Higginson, J.S.; Binder-Macleod, S.A. The relative contribution of ankle moment and trailing limb angle to propulsive force during gait. *Hum. Mov. Sci.* **2015**, *39*, 212–221. [CrossRef]

58. Wagner, D.R. *ACSM's Resource Manual for Guidelines for Exercise Testing and Prescription*, 7th ed.; Kluwer, W., Ed.; 2014. Available online: https://shop.lww.com/ (accessed on 1 May 2021).
59. Sánchez, N.; Simha, S.N.; Donelan, J.M.; James, M.; Finley, J.M. Taking advantage of external mechanical work to reduce metabolic cost: The mechanics and energetics of split-belt treadmill walking. *J. Physiol.* **2019**, *597*, 4053–4068. [CrossRef] [PubMed]
60. Roper, J.A.; Stegemöller, E.L.; Tillman, M.D.; Hass, C.J. Oxygen consumption, oxygen cost, heart rate, and perceived effort during split-belt treadmill walking in young healthy adults. *Eur. J. Appl. Physiol.* **2013**, *113*, 729–734. [CrossRef] [PubMed]
61. Sombric, C.J.; Torres-Oviedo, G. Augmenting propulsion demands during split-belt walking increases locomotor adaptation of asymmetric step lengths. *J. Neuroeng. Rehabil.* **2020**, *17*. [CrossRef] [PubMed]
62. Stenum, J.; Choi, J.T. Step time asymmetry but not step length asymmetry is adapted to optimize energy cost of split-belt treadmill walking. *J. Physiol.* **2020**, *598*, 4063–4078. [CrossRef] [PubMed]