Velocity of the Body Center of Mass During Walking on Split-Belt Treadmill

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Abstract: Walking on split-belt treadmills (each belt rotating at a different velocity) has inspired a growing number of researchers to study gait adaptation and rehabilitation. An overlooked peculiarity of this artificial form of gait is that the mean velocity adopted by the participant, considered as a whole system represented by the body Center of Mass, can be different from the mean velocity of the two belts. Twelve healthy adults (21–34 yrs) were requested to walk for 15 mins on a treadmill with belts rotating at 0.4 and 1.2 m sec^{-1}, respectively (mean = 0.8 m sec^{-1}). Each belt was supported by four 3-dimensional force sensors. For each participant, six strides were analyzed during the 1st and the 15th minute of the trial. The mean Center of Mass velocity was computed as the sum of the velocities of each belt weighted by the percentage of time during which the resulting forces, underlying the accelerations of the Center of Mass, originated from each belt. Across early and late observations, the median Center of Mass velocities were 0.72 and 0.67 m sec^{-1}, respectively ($P < 0.05$). Therefore, the real velocity of the Center of Mass and its time course should be individually assessed when studying walking on split-belt treadmills.

Key Words: Split-Belt Treadmill, Walking, Center of Mass, Rehabilitation

Split-belt treadmills are becoming popular within physiology and rehabilitation research. A simple PubMed search gives back 21 articles in the 2000–2009 decade versus 176 articles in the 2010–2019 decade ("Walking"[Mesh] AND "treadmill" AND "split belt," accessed November 1, 2020). In these instruments, the two parallel belts can rotate at different velocities. Humans adapt in a few strides to this unusual condition. These treadmills were initially proposed for studying the neural mechanisms subtending the adaptation to "split" walking.[1–3] Recently, this paradigm was increasingly applied to the study of walking in people affected by different motor impairments, ranging from hemispheric[4] and cerebellar stroke[5] to Parkinson disease.[6] A therapeutic application of split walking was also proposed based on the observation that claudication can be attenuated by manipulating belts' velocities.[7–12]

This article aims at highlighting that a critical issue was overlooked in this research, namely, the accurate measurement of the mean forward velocity adopted by the participant considered as a whole system, represented by its Center of Mass (CoM). All walking phenomena are velocity-dependent, including the mechanical and neural events characterizing the motion of the lower limbs. The mean velocity of the CoM may differ from the mean velocity of the treadmill belts: this is counterintuitive and is contrary to the assumptions in literature.[13–15] Ignoring this difference prevents a correct clinical interpretation of the asymmetries observed after manipulation of the belts' velocities.

The CoM is a virtual point that can move through and outside the body. The mechanical energy changes (and hence the displacements) of the CoM can be obtained by analyzing the resulting ground reaction forces (GRFs), provided that entire strides are performed on a force-sensorized surface, and the mean velocity of the CoM is known ("double integration," "Newtonian," or Cavagna's method).[16,17] Moreover, the point of application (POA), module, and 3-dimensional orientation of the resulting GRF can be easily computed.[18] The idea is that the CoM can be considered as accelerating or decelerating forward with respect to the forward velocity of the belt where the POA lies. The mean CoM velocity can thus be obtained by summating the velocities of the two belts, each weighted by the percentage of stride time the POA of the GRF originates from each belt.

METHODS

Participants

Twelve healthy adults (7 women) with no history of neurological or orthopedic impairment were enrolled (see Table 1 for demographic data).

Instrumentation

A force-sensorized split-belt (Model ADAL-3D-F-COP-Mz; Medical Development, Tecmachine Hef, Andrézieux Bouthéon, France) was used. The force sampling rate was 250 Hz. Further technical details on the instrument were published elsewhere.[16]

Testing Procedure

Familiarization

The participants wore a t-shirt, short pants, and light gym shoes. They were requested to walk freely on a treadmill with

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TABLE 1. Demographic data of participants ($N = 12$)

Sex, female/male	7:5
Age, mean (SD), yr	27.2 (4.4)
Height, mean (SD), cm	172.3 (7.3)
Weight, mean (SD), kg	66.7 (10.6)
Dominant lower limb, right/left	11:1

belts rotating at the same velocity ("tied"), increasing from 0.4 to 1.2 m sec^{-1} in 0.2-m sec^{-1} increments. Speed increments were applied every 30 secs after a verbal warning. Further details are provided elsewhere.[17]

Testing

The participants started the testing session walking in tied condition at 0.4 m sec^{-1}. After 30 secs, the velocity of one belt, the one under the dominant lower limb,[18] was increased to 1.2 m sec^{-1}. The step initiation was marked by the vertical force exceeding 30 N.[16] Participants had to walk for 15 mins on the treadmill in split condition. Two series of six subsequent strides were considered during the testing period, that is, the second (strides 7–12, tagged "early") and the last (tagged "late") series.

Algorithms

Ground reaction forces resulting from forces recorded by sensors under both belts were computed. The belt over which the POA was located could easily be identified from the horizontal coordinate of the POA itself. During a whole stride, the mean velocity of the CoM was obtained from summating the velocities of the two belts, each weighted by the percentage of time during which the GRF originated from the corresponding belt.

The concept is formalized by Eq. 1. For each single gait cycle (stride):

$$\overline{V}\text{CoM}_{\text{split}} = \left[\overline{V}_{\text{slow}} \frac{t_{\text{slow}}}{t_{\text{stride}}} \right] + \left[\overline{V}_{\text{fast}} \frac{t_{\text{fast}}}{t_{\text{stride}}} \right] \qquad [\text{Eq.1}]$$

where

- $\overline{V}\text{CoM}_{\text{split}}$ (in meters per second) is the weighted mean forward velocity of the CoM during a single stride in split-belt walking.
- V_{slow} and V_{fast} (in meters per second) are the (known and constant) forward velocities of the slower and faster treadmill belts, respectively.
- t_{slow} and t_{fast} (in seconds) are the time intervals during which the POA originates from the slower or the faster belt, respectively, during a given stride.
- t_{stride} (in seconds) is the whole stride duration ($= t_{\text{slow}} + t_{\text{fast}}$).

For the mean values of $\overline{V}\text{CoM}_{\text{split}}$ across six strides in a single participant and across six strides per 12 participants ($N = 72$), the notations $\overline{V}\text{CoM}_{\text{split,6strides}}$ and $\overline{V}\text{CoM}_{\text{split,all}}$ are adopted, respectively. When medians, rather than means, are more appropriate as indexes of central tendency, the notations $\overline{V}^{\text{MED}}\text{CoM}_{\text{split,6strides}}$ and $\overline{V}^{\text{MED}}\text{CoM}_{\text{split,all}}$ are adopted.

Statistics

The normality of distributions was tested through Shapiro-Wilk test. Statistics were based on means (standard deviations [SD]) and 95% confidence intervals for normally distributed variables, as well as medians (25th–75th percentiles) otherwise. Inferential statistics on changes between time points were based on repeated analysis of variance (ANOVA) of data complying with the requirement of normal symmetric and isoscedastic distribution and Friedman ANOVA otherwise. In case of significant ANOVA a Tukey's post hoc test was applied. As an index of test-retest reliability, the intraclass correlation coefficients (ICCs) were computed.[19] The ICC$_{2,6}$ model was adopted ("6" stands for the six strides averaged by each participant). Where the ANOVA assumptions did not hold, Kendall τ on ranks was computed. The significance level was set at a P value of 0.05.

Software

Force data were computed through algorithms developed within the SMART software suite (BTS srl, Milan, Italy). Statistic computations and graphics were performed using STATA (Version 14 SE; STATA Corp, College Station, TX) and SigmaPlot (Version 14.0; Systat Software, Inc, San Jose, CA).
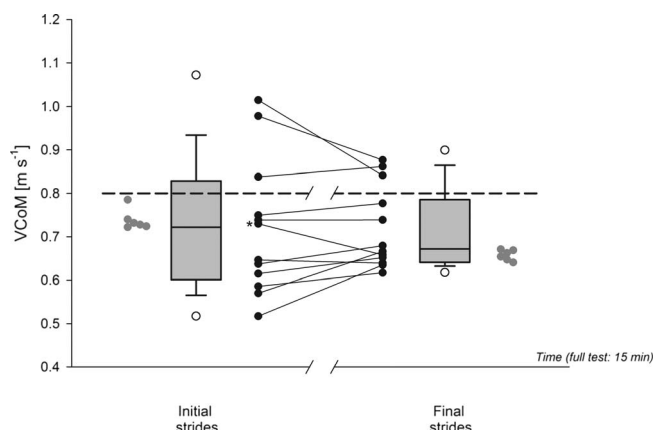

FIGURE 1. The ordinate gives the *V*CoM of the 12 participants (Table 1) walking for 15 mins (abscissa) on a force-sensorized split-belt treadmill with the two belts running at 0.4 and 1.2 m sec^{-1}. Gray circles give the values ($\overline{V}\text{CoM}_{\text{split}}$; see text) of the six early strides (leftmost cluster of symbols) and the six late strides (rightmost cluster of symbols) in a representative participant (male, 26 yrs, 1.76 m, 68 kg). The box plots summarize the distribution of $\overline{V}\text{CoM}_{\text{split,all}}$ of all observations (six strides per 12 participants, $N = 72$) during the early and late series of strides (left and right boxes, respectively). Each box spans from the 25th to the 75th percentiles of the distribution. Whiskers extend from the 10th to the 90th percentiles. Outliers are presented as empty circles. The black circles give the median values across six strides ($\overline{V}^{\text{MED}}\text{CoM}_{\text{split,6strides}}$; see text) for each of the 12 participants. The asterisk indicates the representative participant. The left and right columns of black circles refer to the early and late series of six strides, respectively. Early and late values for the same participant are connected through straight segments ("spaghetti graph") to give an overview of the test-retest stability of the measurements. The dashed horizontal line indicates the average velocity of the two belts (0.8 m sec^{-1}), running at 0.4 and 1.2 m sec^{-1}, respectively. Numeric values are given in Table 2.

TABLE 2. Numerical summary of results and inferential statistics

A

$N = 72$	Early Strides	Late Strides
$\overline{V}\text{CoM}_{\text{split,all}}$; mean (SD)	0.723 (0.161)	0.721 (0.113)
Shapiro-Wilk, P	0.000	0.000
$\overline{V}^{\text{MED}}\text{CoM}_{\text{split,all}}$; median (25th–75th percentiles)	0.722 (0.60–0.83)	0.672 (0.64–0.78)

B

$\text{ICC}_{2,6}$	0.913	CI 95% = 0.698–0.975	$P = 0.000$
MRD (early vs. late series of 6 strides), m sec^{-1}	0.092		

C

Friedman ANOVA ($N = 72$)	P
Time points, $\overline{V}\text{CoM}_{\text{split}}$	0.000
Participants	0.256
Time*participants	0.401

D

Friedman ANOVA ($N = 12$)	P	Kendall τ
Time points ($\overline{V}^{\text{MED}}\text{CoM}_{\text{split,6strides}}$)	0.041	0.923

A, Overall means (SD) of $\overline{V}\text{CoM}_{\text{split}}$ values are given for both early and late stride series. The Shapiro-Wilk test suggests significant deviation from normality of distributions; therefore, medians of $\overline{V}\text{CoM}_{\text{split}}$ values are also provided. B, Test-retest reliability assessment: ICC ($\text{ICC}_{2,6}$ model) and the related MRD estimated as per Tesio.[19] C, Friedman ANOVA on $\overline{V}\text{CoM}_{\text{split,all}}$ values. D, Friedman ANOVA and rank agreement (Kendall τ) of individual $\overline{V}^{\text{MED}}\text{CoM}_{\text{split,6strides}}$ values.

CI, confidence interval; MRD, minimal real difference.

Ethics

All participants provided written informed consent. The local ethics committee of the institution approved the study (Project Code 24C8021_2018).

RESULTS

Demographic information on the participants is given in Table 1.

Figure 1 gives, on the ordinate, the velocity of the body's CoM (VCoM) during the 15-min walking trial. It was noted that in none of the participants the CoM did travel forward at exactly the median (equal to the mean) treadmill velocity, that is, 0.8 m sec^{-1}. In 9 of the 12 participants, $\overline{V}^{\text{MED}}\text{CoM}_{\text{split,6strides}}$ was lower than this velocity. A typical regression toward the mean[20] was observed between the two time points. Nevertheless, during the 15-min trial, no participants "crossed" the median treadmill velocity, thus changing their early "choice" for a velocity lower or, respectively, higher than 0.8 m sec^{-1}.

Table 2 provides a summary of the results and inferential statistics.

At both time points, the 95% confidence intervals of $\overline{V}\text{CoM}_{\text{split}}$ remained below the mean velocity between the two belts. The same held for the median CoM velocity of the sample. The mean velocity of the CoM was stable at the beginning and the end of the 15-min walking trial. This consideration holds both for the sample mean and for individual values (the minimal real difference was never attained),

provided that a normal distribution is assumed. This assumption was a weak one. Nonparametric (Friedman) ANOVA shows that the group median decreased significantly between early and late measurements. The individual participants maintained their rank ordering between the two time points (see Kendall τ, consistent with the high ICC value).

DISCUSSION

The results of this study provide evidence that the actual mean velocity of the body system on split-belt treadmills may not correspond to the mean velocity between the two belts, and it can also change in the same walking trial. At a group level, the median CoM velocity is lower than the mean/median velocity between the two belts, and it tends to become even slower at the end of a 15-min trial. However, care must be taken in aggregating data across strides and participants. The settings of the belts' velocities, in themselves invariant, may be associated with changes of $\overline{V}\text{CoM}_{\text{split}}$ from stride to stride and along successive strides during the same walking trial. Moreover, different participants may show different $\overline{V}\text{CoM}_{\text{split,6strides}}$ values. It is not simply a matter of keeping the body midline right or left of the center line, but one of forces exerted against each belt.

Therefore, in split-gait studies, force-sensorized treadmills should always be adopted. The motion of the CoM can be analyzed through "indirect" methods based on kinematic analysis of the body segments (usually through optoelectronic "capturing" of retroreflective skin markers) as per anthropometric

modeling.[21] During ground walking or traditional treadmill walking, this method has the advantage of locating the CoM with respect to the body segments; however, in "split" treadmill walking, the mean velocity of the CoM cannot be assumed to equate the mean velocities between the belts.

The behavior of participants seems time-dependent. Between-participant variance seems to decrease along the trial. This may reflect something more than simply chance-determined regression toward the mean; rather, it might reflect a form of individual adaptation, mimicking pure random changes at a group level. For instance, overconfidence on the dominant limb might foster a higher velocity, whereas fear of falling or tendency to save muscle power might foster a lower velocity, both behaviors being attenuated by practice. Therefore, the velocity of walking needs to be determined individually, preferably even stride by stride.

The present study is based on a small sample. Notwithstanding, it seems sufficient to direct attention to the problem of the CoM velocity during "split" walking. This finding holds relevance from both research and clinical standpoints. As a hint to physiological research, when segmental motions and their adaptation to this unusual form of locomotion were studied, split-belt trials were often compared with baseline trials at velocities equal to the mean velocity between the two belts.[13–15,17] This error can now be avoided. In addition, the scope of research might now correctly extend beyond the kinematics and dynamics of the lower limbs. Research might now embrace metabolic and ergometric variables of the body system, thanks to estimates of kinetic energy based on CoM velocity. One should consider that errors in estimates of velocity generate squared errors in estimates of kinetic energy.[23] As a hint to rehabilitation research, it has been already highlighted that an efficient translation of the CoM (across the whole stride, hence per unit distance) may coexist with different impairments.[24,25] This paradoxical efficiency may help in deciding whether, and by how much, focal alterations represent an adaptation to, rather than a direct source of, walking abnormalities. In both cases, a high efficiency of the COM transfer possibly prevents recovery, either spontaneous or based on rehabilitation. A sort of "acquired/learned non-use" seems to affect the impaired lower limb. Increasing the dynamic requests of walking (i.e., by asking the patient for a higher velocity or to walk uphill) succeeds in obtaining a greater muscle work and power from both lower limbs, so that the acquired dynamic asymmetry, more than unilateral weakness itself, seems to be the invariant constraint of these gaits.[17,22] Split-belt walking paradigms allow a deeper investigation of this intriguing finding, as far as they allow experimental manipulation, not only observation, of the asymmetry. On the diagnostic side, the actual velocity of the CoM (for any given pair of belts' velocities) might become in itself a primary measure of walking performance and an index of improvement after rehabilitation. The propulsive role of either lower limb can emerge from the simultaneous recording of local and CoM power changes. On the therapeutic side, the many divergent exercise paradigms on split-belt treadmills (e.g., impaired lower limb on the slower and/or on the faster belt[10]; different duration and scheduling of "split walking" sessions; imposed, vs. self-selected velocities of the belts) might be compared with respect to their capacity to restore symmetry of the locomotor mechanism as a whole, not only to ameliorate focal joint kinematics and dynamics. The same holds for other potential treatments consistent with the hypothesis of "learned non-use" during asymmetric walking,[22] such as lower limb forced-use exercises,[26] noninvasive brain stimulation,[27] and mirror training.[28]

In these types of studies, walking on tied belts represents the control condition. Is the latter analogous to overground walking? Minor kinematic differences, for the same average velocity, have been outlined in adults.[29,30] These differences seem amenable to a step length shorter (hence to a cadence higher) by no more than 10%, in treadmill compared with ground walking. Step shortening is perhaps caused by a higher cognitive effort[30] required to counteract the conflict between proprioception (signaling motion) and vision (signaling immobility). When height-adjusted, dynamic equivalent velocities (i.e., transformed into the same adimensional Froude number) are compared, the same holds for children aged 5–13 yrs.[31] Cadences above or below the "optimal" one entail a less efficient pendulum-like transfer of energy within the CoM motion (hence, a higher energy expenditure per unit distance), but with impact negligible for changes lower than 50%.[32] For this reason, results on split-belt treadmills can be considered as enlightening the pathophysiology of "natural," overground walking (for a discussion on this topic, see the study by Tesio and Rota[16(p525)]).

Although foreshadowed by the present study, all applications of split-belt treadmills require further research on the velocity of the CoM. For instance, open research questions are as follows: which is the actual velocity of the CoM when various differences between the belts' velocities are imposed? What is the time course of changes in CoM velocity during a walking test? How strong is the after-effect in CoM velocity?

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