Optimizing Prosthetic Gait

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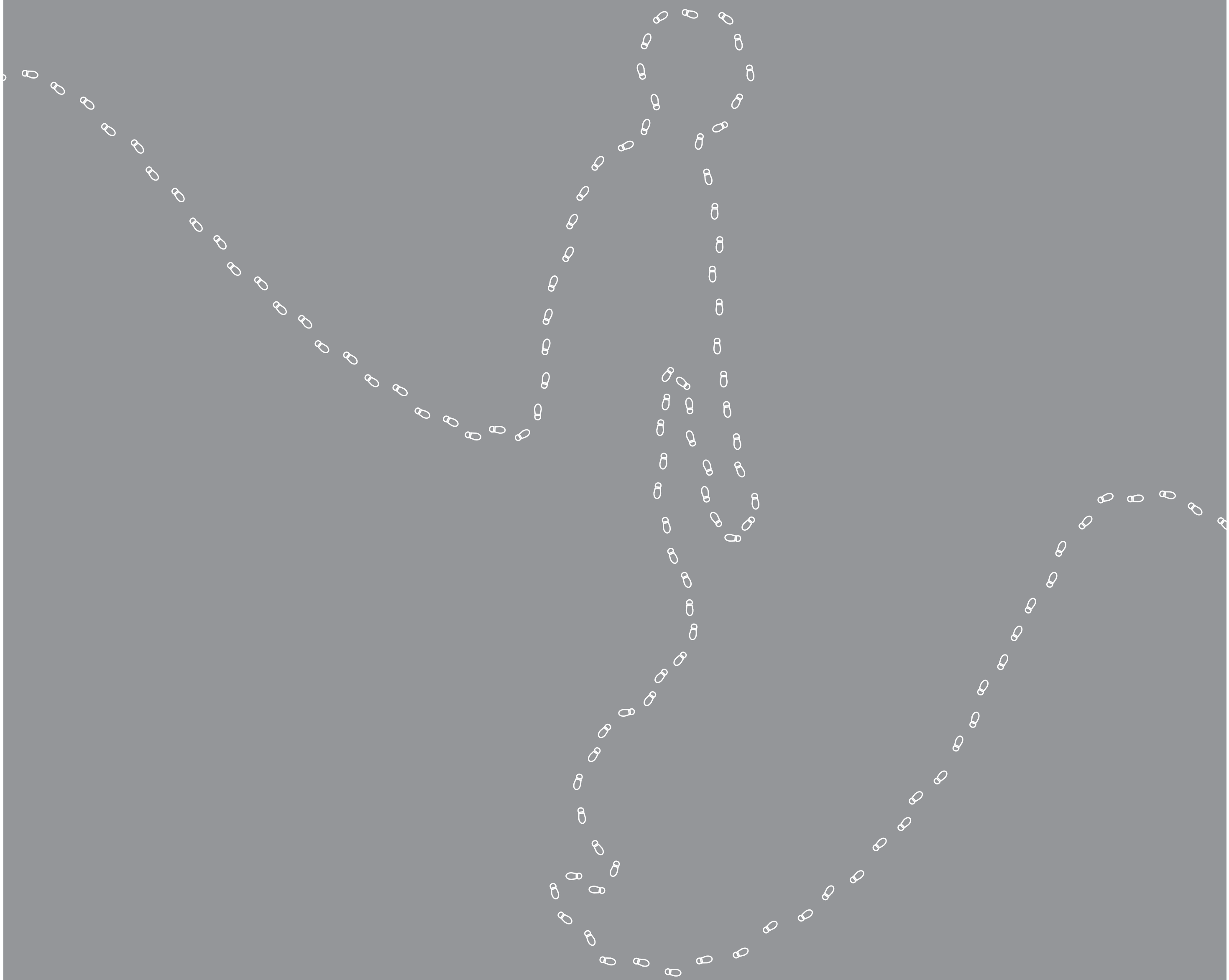
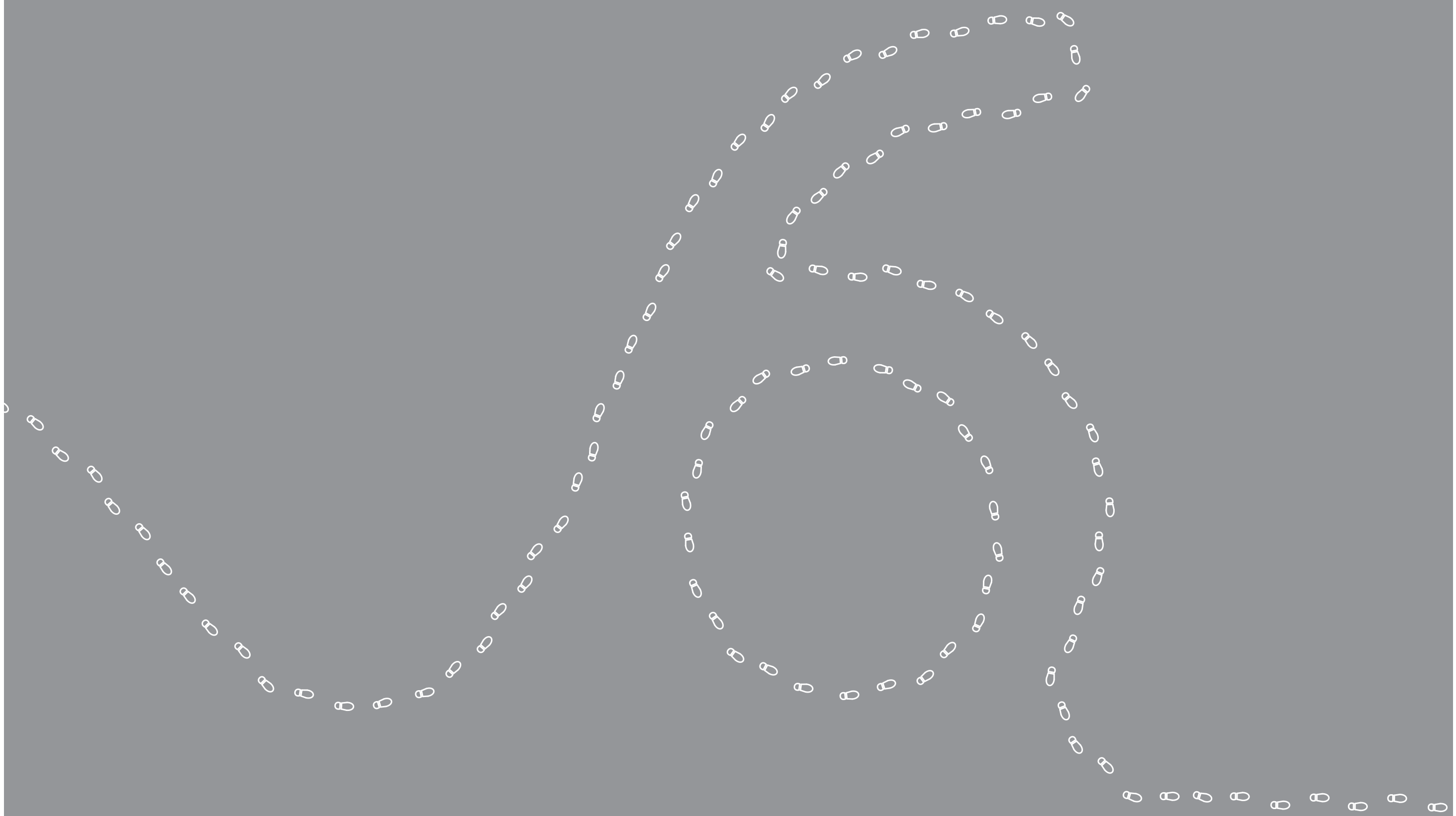


Mind your step: Metabolic energy cost while walking an enforced gait pattern



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Abstract

The energy cost of walking could be attributed to energy related to the walking movement and energy related to balance control. In order to differentiate between both components we investigated the energy cost of walking an enforced step pattern, thereby perturbing balance while the walking movement is preserved. Nine healthy subjects walked three times at comfortable walking speed on an instrumented treadmill. The first trial consisted of unconstrained walking. In the next two trials, subject walked while following a step pattern projected on the treadmill. The steps projected were either composed of the averaged step characteristics (periodic trial), or were an exact copy including the variability of the steps taken while walking unconstrained (variable trial). Metabolic energy cost was assessed and center of pressure profiles were analyzed to determine task performance, and to gain insight into the balance control strategies applied. Results showed that the metabolic energy cost was significantly higher in both the periodic and variable trial (8% and 13%, respectively) compared to unconstrained walking. The variation in center of pressure trajectories during single limb support was higher when a gait pattern was enforced, indicating a more active ankle strategy. The increased metabolic energy cost could originate from increased preparatory muscle activation to ensure proper foot placement and a more active ankle strategy to control for lateral balance. These results entail that metabolic energy cost of walking can be influenced significantly by control strategies that do not necessary alter global gait characteristics.

Introduction

Patients require more metabolic energy to walk at the same speed as their matched healthy controls [225]. This increase has been attributed to an altered and less efficient movement pattern [44, 110, 132, 140], an increased effort for balance control [109, 148] or a combination of both [103, 140, 148]. As early as 1986, Workman and Armstrong [237] described a model which explicitly differentiated between the energy cost associated with balance preservation and the energy cost needed to perform the walking movement. In subsequent years, however, many studies have focused predominantly on alterations in the walking movement in terms of external and internal mechanical work to explain differences in energy cost between normal and pathological gait [44, 132]. In contrast, the energy cost associated with balance preservation has deserved less attention. Yet, research has demonstrated that, even during upright standing, balance perturbations result in higher metabolic energy expenditure [109]. Moreover, it has been shown that metabolic energy cost while walking can be decreased by facilitating balance control by means of an external stabilizing device [41, 57, 161].

Differentiation between the energy cost needed to perform the walking movement and that associated with balance control is less clear than the model of Workman and Armstrong suggests. Frequently, balance perturbations directly lead to adaptations in the walking movement [236], for example, many patients walk with increased step-width to extend the margins of the base of support [57, 102, 128] resulting in a more stable gait. However, the dynamics of this wider gait pattern will also result in increased mechanical work and consequently higher metabolic energy expenditure [54]. The question arises whether this extra metabolic energy should be assigned to the metabolic energy cost related to balance control or to that associated with the walking movement. Since many balance control strategies alter gait characteristics, the energy requirement of alternative mechanisms controlling balance, as for example muscle co-activation [103] and increased local dynamic stability [51], becomes indeterminable.

Information about the energetic demands of these alternative control strategies can be gained by perturbing balance, while restricting alterations in the averaged gait characteristics. Experimentally, this can be accomplished by enforcing the self-selected gait pattern on a treadmill, thereby restricting alteration in global gait characteristics like speed, step-width, step length and dual support time. Enforcing a gait pattern simultaneously imposes a balance perturbation. Normally, humans primarily use a stepping strategy to maintain stability when walking; balance perturbations in one step are corrected in consecutive steps by appropriate placement of the foot [102]. However, when foot positions are enforced, the use of a stepping strategy will consequently be restricted. Therefore, subjects have to revert to an altered, possibly less efficient, balance control strategy. In addition, metabolic energy cost might be influenced by the fact that placing the foot at a predefined location will require more conscious control [7, 59] and associated preparatory muscle activation.

The primary objective of this study was to investigate the influence of walking at an enforced gait pattern on the metabolic energy cost in young healthy subjects. By enforcing a gait pattern balance is perturbed, without allowing global gait characteristics to change. It was hypothesized that enforcing a gait pattern, composed of the averaged self-selected gait pattern, would lead to a substantially higher metabolic energy cost compared to walking unconstrained. The increase in metabolic energy cost was thought to be even higher when an exact copy of the self-selected gait pattern is enforced, since the variability, naturally present when walking, result in a less predictable enforced walking pattern. In addition to the primary objective, gait parameters were measured to determine the accuracy of task performance and center of pressure (COP) profiles were analyzed to gain insights into the balance control strategies applied.

Method

Subjects

Nine subjects with an average age of 27.6 (SD = 9.9) years and a body mass index of 21.3 (SD = 1.8) participated in this study. Subjects were healthy, having no muscular, neurological or visual limitation, and were asked to refrain from drinking coffee and eating large meals at the day of the test. Subjects were informed about the research procedure before they gave informed consent. This study was approved by the local ethics committee.

Equipment

Subjects walked on an instrumented treadmill (C-Mill, ForceLink, Culemborg, The Netherlands). The treadmill was mounted with a force plate (1.5 by 1 m) from which step characteristics could be derived. Data acquisition was performed using a 32 bits A/D converter at 100 Hz. The instrumented treadmill was connected to a projector which could project step patterns onto the walking surface.

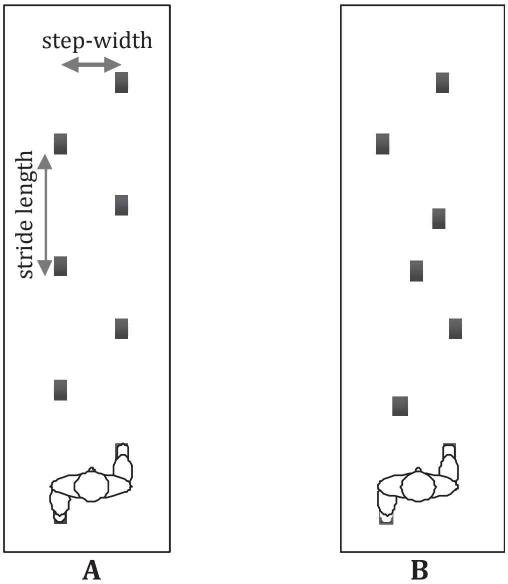

Figure 1. Schematic representation of the enforced walking trials.

The projected steps started three meters in front of the subject and consisted of rectangular shape of 30 by 10 cm. Figure A represents the periodic trial. Projections were based on the averaged values of stride length, spatial asymmetry and step-width while walking unconstrained, the variation is zero. Figure B represents the variable trial. In this trial projections consisted of an exact copy of the steps made while walking unconstrained. Variations in step parameters in this trial were equal to the unconstrained trial.

Walking trials

After a familiarization period on the treadmill, subjects' comfortable walking speed was obtained using the method described by Martin and colleagues (1992) [141]. Thereafter, subjects walked three times 6 min at comfortable self-selected speed separated by a minimum of 10 min rest. The first trial consisted of unconstrained walking. In the next two randomly assigned trials, the gait pattern was enforced by means of a step pattern projected on the treadmill. Each projected step consisted of a yellow rectangular shape, measuring 30 by 10 cm, projected on the black treadmill belt. The projections moved at the same speed as the treadmill and started 3 m in front of the subject, which enabled appropriate anticipation of foot placement. Subjects were asked to place their feet within the projected surface. In one of the enforced trials, subjects were confronted with a gait pattern composed of the averaged stride length, spatial asymmetry and step-width determined using the data collected in the unconstrained trial. The projections in this trial were periodic and contained no variability; therefore, this trial will be referred to as the periodic enforced trial (Figure 1). In the second enforced trial, a gait pattern was projected that was an exact copy of the steps taken during the unconstrained trial. Although on average the same walking pattern was enforced compared to the periodic trial, subjects were now confronted with an amount of variation equal to the variation while walking unconstrained. This trial will be referred to as the variable enforced trial (Figure 1).

During the enforced trials task execution was visually monitored, and stimulating feedback was given in order to redirect attention towards appropriate foot placement whenever subjects showed a decline in task performance.

Data analysis

To ensure that steady state was reached, only data collected during the last minute of each walking trial was used for analysis. Oxygen consumption($\dot{V}O_2$, $ml \cdot min^{-1}$) was measured breath-by-breath using open circuit respirometry (Oxycon delta, Jaeger, Hoechberg, Germany). $\dot{V}O_2$ was converted to metabolic power (P_{met}) using the following equation:

$$P_{met} = (4.960 \cdot RER + 16.040) \cdot (\frac{\dot{V}O_2}{60}) \quad (1)$$

where RER is the respiratory exchange ratio, which in all analyzed trials was below one. P_{met} was normalized for bodyweight and walking speed to obtain metabolic energy cost (C_{met}, $J \cdot kg^{-1} \cdot m^{-1}$).

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Using a local reference frame within the force plate, center of pressure (COP) profiles were collected during the last minute of the walking trial. The profiles resembled a butterfly pattern from which heel contact was defined as the local minima in anterior-posterior direction contra-lateral to the side of interest. Toe-off was determined by identifying a local minimum in the total force that occurred around the local maxima in the COP profile as previously outlined by Roerdink and colleagues (2008) [177].

From known toe-off and heel contact instants, stride length (cm), dual support time (%) and step-width (cm) were calculated. Concretely, stride length was defined as the distance between heel contact and consecutive heel contact of the same leg. Dual support time was expressed as a percentage of the gait cycle when both feet were in contact with the ground. Step-width was defined as the absolute distance in frontal plane of the COP at consecutive periods in the gait cycle when both feet were in mid-stance (Figure 1). Mid-stance was defined as the instant at which COP had travelled half the distance between toe-off and heel contact.

Compliance with the task was monitored by comparing the mean stride length, dual support time and step-width between trials. The accuracy with which the foot was positioned within the projected surface was calculated as the standard deviation around the averaged COP location within the projection in medio-lateral and anterior-posterior direction at mid-stance.

Differences in balance control strategies, applied during single limb support, were analyzed by investigating the COP displacement under the foot [102].

Table 1. Mean (± SD) step parameters.

	unconstrained	periodic	variable
Stride length (cm)	135 (9.6)	135 (9.7)	135 (9.6)
Dual support time (%)	22 (3.4)	23 (3.4)	23 (3.5)
Step-width (cm) *	11 (3.6)	13 (2.9)	11 (2.4)

NOTE. Mean (± SD) for stride length (cm), dual support time (%) and step-width (cm). The significant difference in step-width is denoted using an asterisk (*). Pair wise comparisons of step-width failed to reach significance.

Previous research has shown that a more variable COP displacement in medio-lateral direction during single limb support indicates a more pronounced ankle strategy [102]. The amount of COP displacement under the foot was determined as the standard deviation in medio-lateral COP position during single limb support. The averaged value over all single support phases during the last minute of data collection was used in the statistical analysis.

Statistics

Data were tested for normality using the Kolmogorov-Smirnov test. All parameters were normally distributed. Statistical analysis to test for differences between trials was performed using repeated measurement ANOVA (SPSS Inc., Chicago, IL, USA), with subjects as between factor and trial as the within factor. Multiple comparison correction was made using a Sidak post hoc test. The consistency between trials in global gait characteristics (stride length, dual support time and step-width) was tested using intra-class correlation coefficient ($ICC_{(3,1)}$) with a two-way mixed model for single measurements [196]. The accuracy of foot placement between the two enforced trials was tested using paired sample *t* test. Significance was set at the 5% level. Effect sizes were calculated using eta squared (η^2) [69].

Results

The metabolic energy cost (C_{met}) was significantly higher in both the periodic and variable walking trial (4.38 ± 0.34 and 4.59 ± 0.45 $J \cdot kg^{-1} \cdot m^{-1}$, respectively) compared to unconstrained walking (4.05 ± 0.35 $J \cdot kg^{-1} \cdot m^{-1}$, $p < .001$; Figure 2). Effect sizes range between 0.42 and 0.62 indicating a medium to large effect (Figure 2) [69]. The difference between the periodic and variable trial failed to reach significance ($p = .09$).

Mean stride length and dual support time did not differ between trials ($p = .724$ and $p = .078$, respectively). This in contrast to mean step-width ($p = .017$). Step-

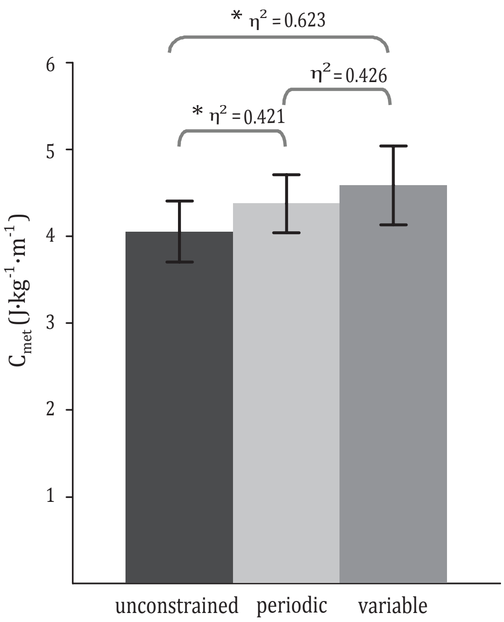

Figure 2. Metabolic energy cost.

Metabolic energy cost while walking unconstrained and at the periodic and variable enforced gait patterns. The metabolic energy cost (mean ± SD) is expressed in joule per kilogram per meter distance travelled ($J \cdot kg^{-1} \cdot m^{-1}$). The effect sizes ($\eta^2$) are noted for all comparisons. All effect sizes indicate a medium or large effect. Statistically significant difference ($p < .05$) between trials are denoted using an asterisk (*).

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width seemed to be larger in the periodic trial, although pair wise comparison failed to reach significance (p = .091 for periodic versus unconstrained and p = .083 for periodic versus variable; Table 1). Further analysis of the consistency over trials revealed that single measurement $ICC_{(3,1)}$ [196] (95% CI) was 0.99 (0.98 - 0.99) for stride length, 0.94 (0.83 - 0.99) for dual support and 0.76 (0.45 to 0.93) for step-width.

The accuracy of foot placement was analyzed by calculating the amount of deviation around the averaged COP location within the projection at mid-stance. Larger deviations were found in medio-lateral direction in the variable trial compared to the periodic trial (right foot; p = .013, left foot; p = .022). In anterior-posterior direction no differences were found between trials (right foot; p = .432, left foot; p = .344; Figure 3).

The variation in COP displacement under the foot during single limb support differed between trials (left side; p = .002, right side; p = .004). Figure 4a illustrates the medio-lateral COP displacement under the left foot during single limb support for a number of consecutive steps of a typical subject. The overall mean variation (cm) in consecutive steps is illustrated in Figure 4b. The periodic and variable trials both showed a higher amount of variation (1.14 and 1.21 cm, respectively) in the medio-lateral direction when compared to the unconstrained trial (0.84 cm). However, differences between the unconstrained and periodic trial failed to reach the set level of significance (left side; p = .076, right side; p = .057), although a trend can be noted.

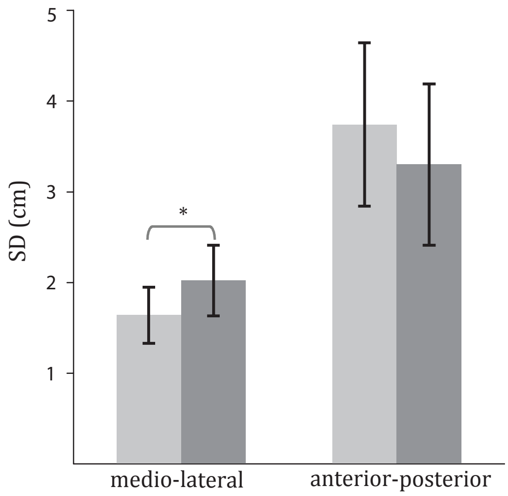

Figure 3.Task compliance.

Task compliance measured using the standard deviation of COP position at mid-stance (cm), relative to the averaged position within the projected surface, in both medio-lateral and anterior-posterior direction. The light bar represents the periodic trial and the dark bar the variable trial. Values are mean and ± SD, only data of the left step are presented. Statistically significant difference ($p < .05$) between trials are denoted using an asterisk (*).

Discussion

This study investigated whether balance and movement control can contribute significantly to the metabolic energy cost of walking, independent from alterations in global gait characteristics. By enforcing subjects to walk at their self-selected gait pattern, alterations in the global gait characteristics were restricted while balance was perturbed. From the two enforced trials, the variable trial was hypothesized to elicit the largest effect on metabolic energy cost. In accordance with our hypothesis, C_{met} increased when a walking pattern was enforced compared to walking unconstrained. This increase reached 8% in the periodic trial and 13% in the variable trial. A similar (14%) increase in C_{met} was previously found by Donelan and colleagues (2001) when subjects were forced to follow their self selected step-width [54]. An increase in metabolic cost of more than 10% is commonly regarded as a relevant change [13, 190, 206]. Although this value is partly based on subjective clinical opinion, it implies that our experimental manipulation elicited a substantial effect during the variable trial, and that the periodic trial failed to reach this criterion.

Comparing the enforced trials

It was hypothesized that the variable trial would elicit the highest effect on metabolic energy cost. The highest energy cost was indeed found in the variable trial, however, differences between the periodic and variable trial failed to reach the set level of significance. Calculation of the effect size between the periodic and variable trial ($r = 0.67$) revealed that 45% of the total variance could be attributed to the task [69]. Due to the small sample size, the effect size might give

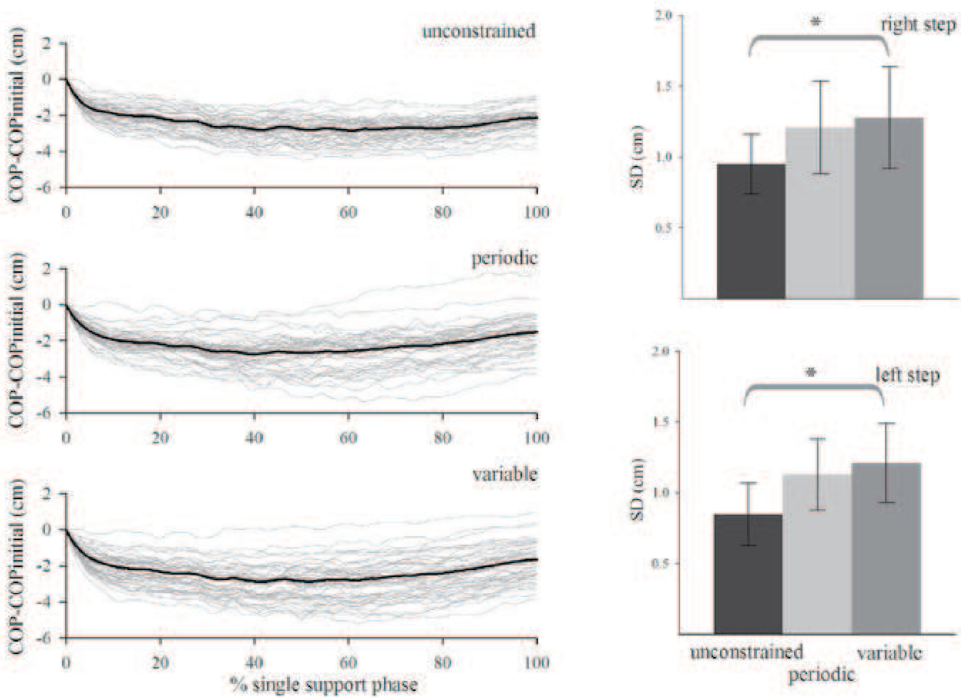

Figure 4. COP trajectories.

(A) COP trajectories in medio-lateral direction (cm) during single support phase of the left leg of a typical subject. The initial position is set at zero to illustrate variation in the trajectory of COP between consecutive steps, during single limb support. The x-axis is normalized to represent the % of single support phase. The light colored lines represent COP trajectory of a number of single support phases during the last minute of data collection. The dark line represents the averaged trajectory of COP. Negative values indicate a position lateral to the initial position of COP at the start of single limb support. **(B)** Averaged values of COP displacement during single limb support for all three trials. The amount of displacement was determined by calculating the standard deviation of medio-lateral COP position (cm) during the single limb support phase. Both right and left steps are depicted. Variability of the medio-lateral COP trajectories was higher in the enforced trials compared to unconstrained walking. Statistically significant difference ($p < .05$) between trials are denoted using an asterisk (*).

a more objective measure of the importance of the observed effect. Additionally, analysis of the COP position at mid-stance showed more deviation in medio-lateral direction, with respect to the averaged COP position within the projected surface, in the variable trial compared to the periodic trial (Figure 3). This finding indicates that the error in foot placement was larger in the variable trial and further supports the notion that during the variable trial balance was perturbed more severely, consequently resulting in a higher metabolic load compared to the periodic trial.

Compliance with the task

A high level of compliance with the task is essential for a valid interpretation of the results. Subjects were able to accurately position the foot within the projected surface. The averaged COP position at mid-stance was well within the projection and deviated only to a small extent between subsequent steps (medio-lateral = 1.8 cm, anterior-posterior = 3.6 cm; Figure 3).

Considering that the enforced trials were based on the unconstrained walking trial, averaged step parameters should be equal to the values when walking unconstrained. This was true for the stride length and dual support time. Step-width was wider in the periodic trial compared to the unconstrained and variable trial, although, pair wise comparison failed to reach statistical significance and the analysis of the consistency over trials revealed a high consistency ($ICC_{(3,1)}$ = 0.76). The wider step-width could have amplified the found increase in metabolic energy cost. However, using the equation proposed by Donelan and colleagues (2001) [54] only 1.2% of the increase in metabolic cost found in the periodic trial could be explained by the wider step-width. The small deviation in COP location within the projected surface at mid-stance and the comparable mean gait parameters seem to justify the conclusion that subjects complied with the task sufficiently.

Control strategies

Considering that subjects were able to comply with the task and walked at equal gait patterns over all trials, means that mechanical adaptations in the global walking pattern were limited. Therefore, observed increases in the metabolic energy cost should be related to changes in control strategies other than those altering global gait characteristics. A number of control strategies could have been employed to overcome the balance perturbation due to the enforced gait pattern, and consequently explain the higher oxygen cost found. First, the step pattern was projected well before foot contact, enabling sufficient time for subjects to anticipate on the upcoming step. Therefore, subjects possibly relied to a lesser extent on the passive dynamics of their swing leg but consciously made anticipatory modifications of their leg's trajectory. The precise placement of the foot is mostly regulated by activation of the responsible muscles just prior to foot contact (i.e. at the end of swing phase). This would, in contrast to walking unconstrained, result in more motor cortical involvement [7, 59]. In addition, the precision demands imposed eventuate in more muscle co-activation [193, 235]. These adaptations in motor control and preparatory muscle activity may partly explain the increased metabolic energy cost observed [106]. This notion is substantiated by the fact that the variable (less predictable) trial resulted in the highest metabolic energy cost. However, further research incorporating electromyography recordings are warranted to substantiate these conjectures.

Secondly, human gait is laterally unstable [57, 128]. Normally, lateral stability is ensured by accurate placement of the foot (i.e. a stepping strategy) [8, 102]. Due to the restrictions in foot placement a stepping strategy could not be used. Other strategies were needed, as for example a lateral ankle strategy. The lateral ankle strategy requires an active feedback loop and presumably more energy (i.e. is less efficient) than the stepping strategy [8, 102]. Analysis of COP displacement during single limb support showed more variation in medio-lateral COP patterns during single limb support in the enforced trials compared to unconstrained walking, this effect was largest for the variable trial. This result supports the idea that, after positioning of the foot, a greater amount of fine tuning was required by means of an ankle strategy. In addition to an increased use of the ankle strategy, alterations in hip ad-/adduction and movements of the arms might have contributed to the measured increase in metabolic energy cost; however, these latter responses could not be assessed in the current experimental set-up.

Conclusion

Walking at an enforced gait pattern elicited a substantially higher metabolic energy cost. The greater variation in COP trajectories during single limb stance supports the idea that subjects relied more on an ankle strategy to overcome the inability to use the more efficient stepping strategy. This, in addition to altered preparatory muscle activation, is thought to have contributed to the increased metabolic energy cost found.

Obtained results have both a clinical and experimental implication. As for the clinical implication results demonstrate that increased metabolic energy cost in patients could, in addition to an altered gait pattern, originate from changes in balance control strategies that do not necessarily alter global gait characteristics. Experimentally, results underline that metabolic energy cost needs to be interpreted with caution when a gait pattern is enforced; the enforcement in itself could lead to alteration in the metabolic energy cost.