

Effects of Added Inertia and Body Weight Support on Lateral Balance Control During Walking

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Abstract—A robot-driven gait orthosis which allows balance training during gait would further enhance the capabilities of robotic treadmill training in gait rehabilitation. In this paper, additional mass is attached to walking able-bodied subjects to simulate the effects of additional inertia and body weight support on the lateral balance task. The combination of additional inertia and body weight support led to reduced step widths, suggesting a stabilising effect which may reduce the challenge of the lateral balance task.

I. INTRODUCTION

Robot-assisted gait training can be used in rehabilitation of neurologically impaired persons who have suffered a stroke or spinal cord injury [1], [2], [3]. The robotic assistance can be used to provide extra support, especially for those phases of gait where an impairment could otherwise prevent unaided walking. The robotic support of driven-gait devices can be advantageous; therapist-assisted training is physically demanding and the duration of the therapy can consequently be limited, and walking may even be impossible due to spasticity [4]. Robot-driven gait orthoses can not only allow training of longer duration, but also produce a more consistent locomotive pattern than could be realised via manual assistance from a therapist [5], [6]. Furthermore, the instrumentation of the robotic devices allows accurate assessments of, for example, muscle strength and spasticity [6].

A number of therapeutic benefits of robot-assisted gait training have been reported. For instance, effects from training in the robotic gait orthosis Lokomat [7] include improvements in gait speed [2] and muscle tone [2] of stroke patients, and increases in gait speed [3], endurance [3] and joint range of motion [8] have been demonstrated for spinal cord injured patients. Similarly, improvements have been shown for the gait trainer LOPES [9] in terms of joint range of motion and walking speed in stroke patients [10], and walking in the Gait Trainer GT I [11] was shown to improve the independent walking abilities of stroke patients [12].

Despite the therapeutic benefits and potential advantages of robot-driven gait systems, recent data have demonstrated greater improvements through conventional, therapist-assisted training over robotic-based training in moderately to severely impaired patients [13]. Part of the reduced effectiveness of Lokomat training has been postulated to arise from the constraints on the pelvis imposed by the robot [13], causing

changes in the kinematics during gait [14], [15]. Passivity, where the patient shows a lack of active participation in the movement and over-reliance on the machine's assistance, may also be detrimental to motor learning where active participation is beneficial [16], and also to cardiovascular training aspects [17].

Maintaining balance in the frontal plane is, along with providing propulsion and support against gravity, one of the major tasks of walking. During walking, subjects must actively control balance in the frontal plane [18]. Conversely, balance in the sagittal plane is thought to be passively stable and therefore does not possess as strong an active control element as seen in the frontal plane [19]. Control of lateral balance is largely achieved by predicting the future position of the centre of mass and adjusting subsequent foot placement [20], [21]. Movements of the centre of mass during walking may be seen as having step-to-step stability, in contrast to the continuously stable scenario of quiet standing [18].

Some devices such as the Lokomat restrict movement to the sagittal plane [22], make weight shifting from one leg to the other difficult [14], and as a result do not allow training of balance in the frontal plane. Thus, it has recently been proposed to enhance the robotic rehabilitation technology by incorporating balance training into gait therapy with the rehabilitation robot Lokomat by the addition of further degrees of freedom of the pelvis and legs, which would have the additional effect of permitting a more natural gait pattern.

In investigating the control of balance in the frontal plane, it is useful to measure step width since this is recognised as a key parameter of lateral balance. Very narrow or wide steps are metabolically costly, and there is a preferred step width for human walking at which the metabolic cost is minimised [23]. Results have suggested that external factors which artificially stabilise the body lead to a reduction in step width [24], [25].

Body weight support systems and robotic actuators are common features of robotic gait technology, and are required both for help with weight bearing and for propulsion throughout the gait cycle. These features could, however, potentially result in a human-robot system with quite different dynamics as would be seen in the walking human alone. The device will likely impose additional inertia, friction and weight, and these factors - inertia in particular - can only partially be compensated for by control [26], [27]. The body weight support system, in

addition to its primary function of providing vertical support against gravity, could also provide additional support in other directions, reducing the challenge and effort needed from the subject for balance and postural control during walking.

Using measurements of step width, the work presented here investigates whether the lateral balance task is affected by additional inertia and also by horizontal forces from the body weight support system.

II. METHODS

A. Experimental Procedures

Five able-bodied subjects whose characteristics are shown in Table I walked on a Mercury treadmill (h/p cosmos, Nussdorf-Traunstein, Germany). This treadmill is equipped with 8 force sensors embedded in 2 individual plates (at the front and back of the treadmill), allowing computation of the centre of pressure (CoP) of each step [28].

Subject	Age	Gender	Mass (kg)	Height (cm)	Leg (cm)
A	26	M	76	190	98
B	26	F	54	164	88
C	30	M	72	186	97
D	26	F	47	164	86
E	29	M	64	181	94

TABLE I
SUBJECT CHARACTERISTICS.

Subjects walked with additional mass attached to their waists using diving weights. Walking was performed at 3 different mass conditions of 0 kg, 14kg and 28kg. 28kg was found to be the maximum mass that could be securely fastened to all subjects using the diving weight equipment.

Furthermore, 2 different treadmill speeds of 3km/h and 5km/h were used. The lateral position of the centre of mass could be approximated as a sinusoidal function of the form $a \sin(\omega t + p)$ where a is the amplitude and p the phase, and so the corresponding acceleration would be equal to $-\omega^2 \sin(\omega t + p)$. Therefore, peak inertial forces would be expected to vary with the squared frequency ω^2 and consequently, if added inertia were influential in the lateral stability task, an effect of step cadence on the step width should be observed. Therefore, different walking speeds were incorporated into the tests to permit such an effect.

The 6 walking conditions for each subject are summarised in Table II. The order of these conditions was randomised for each person.

The static weight of the additional mass was compensated for using the body weight support system Levi (Hocoma AG, Volketswil, Switzerland), so that the overall effect was an increase in inertia, but not static weight, of the walking subject. The experimental set-up is illustrated in Fig 1.

The body weight support system will tend to produce a horizontal force, since deviations from the vertical position of the cable will produce a lateral force component towards the centreline, as shown in Fig 2. With the force acting in opposite direction to the subject's lateral movement, the body weight

Condition	Added Mass (kg)	Walking Speed (km/h)
C1	0	3
C2	14	3
C3	28	3
C4	0	5
C5	14	5
C6	28	5

TABLE II
DIFFERENT CONDITIONS OF ADDITIONAL MASS AND WALKING SPEEDS USED IN THE TESTS.

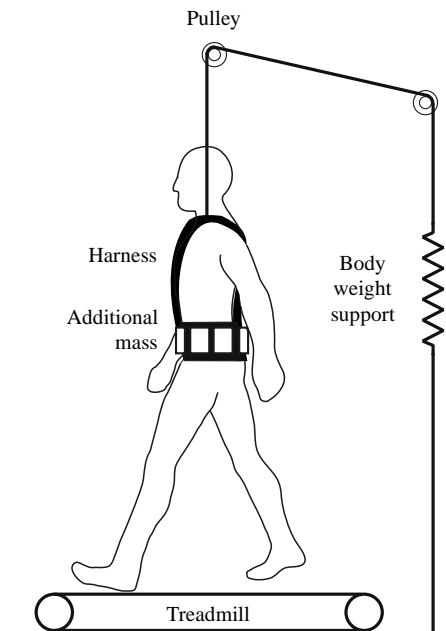


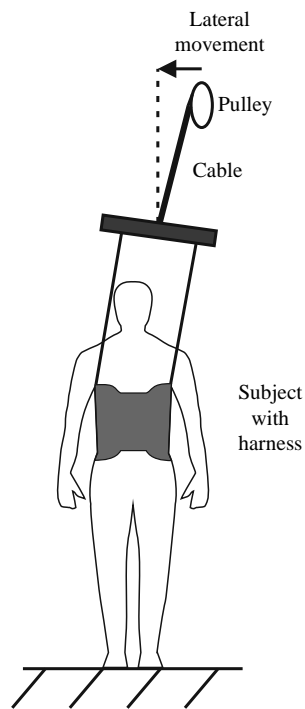
Fig. 1. Experimental setup showing treadmill, body weight support system and additional mass attached to the subject.

support system acts as a spring which provides assistance in the lateral balance task.

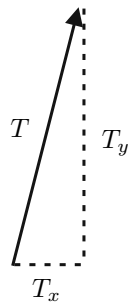
B. Data Processing and Statistical Analysis

Data from the front plate of the treadmill were used to calculate the CoP for the first portion of each step. The average lateral position of the CoP was calculated, and differences in lateral positions between left and right steps were then used to calculate a vector of step widths. This subsequently allowed the overall mean step width, and also the mean step cadence to be computed for that condition. The process of calculating the step widths is illustrated in Fig 3.

The effects of added mass and walking velocity on step width were tested for using a 2-way ANOVA. The data was checked for Gaussianity by visual inspection. Note that the mean step widths of each subject were normalised using the leg length of that subject. Moreover, the effects of speed and loading on step cadence were also tested using a 2-way ANOVA. The significance level was set at 5%.



(a) Subject's lateral movement induces an angle in the cable.



(b) Lateral component of tension.

Fig. 2. Lateral subject movement induces an angle in the body weight support cable, causing a stabilising force to be applied to the subject in the lateral direction.

III. RESULTS

Mean step width (across subjects and walking speeds) was 20.5% lower with a load of 28kg as compared with no load. Fig 4 shows an example of the mean trajectory of the centre of pressure for 2 loading conditions. It can be seen that the centre of pressure for the heavier loading case lies medial to that of the unloaded scenario, giving a correspondingly smaller mean step width.

The ANOVA on cadence showed that there was a significant effect of treadmill speed ($p \ll 0.01$) but not of the loading condition ($p = 0.87$) on the step cadence. Fig 5 shows the step cadences at different loading and treadmill speed conditions in box plot form.

Box plots for the step width data are given in Fig 6. The ANOVA on step width indicated a significant effect of loading

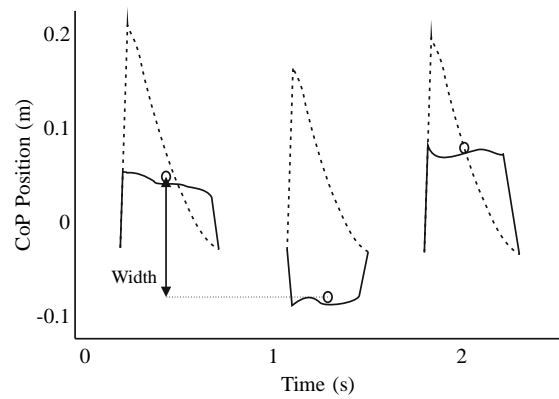


Fig. 3. Peaks in the longitudinal position of the CoP (dashed) indicate the beginning of each step, allowing averaged lateral positions (open circles) to be calculated from the lateral CoP position (solid lines), giving the step widths.

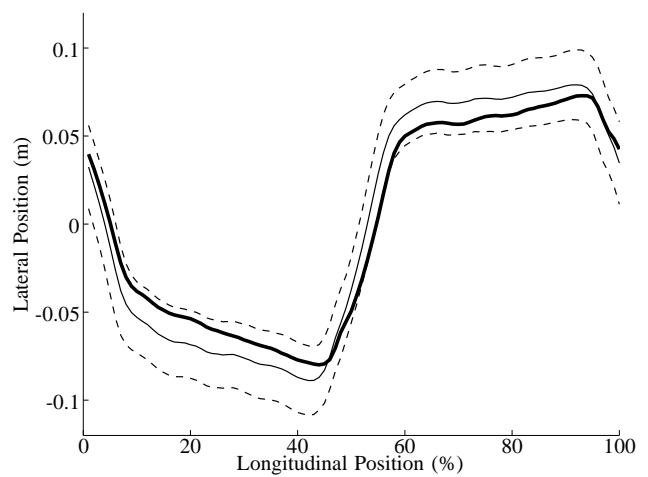


Fig. 4. The mean positions of the centres of pressure are depicted by the solid and lines for subject E at 3km/h. The thin solid line is used for the 0kg case and the thick solid line for the 28kg case. The dashed lines represent standard deviations in the lateral position in the 0kg case.

condition on the (normalised) step widths ($p = 0.015$) but not of walking speed ($p = 0.781$), and no significant interaction between the 2 factors ($p = 0.493$). In general, subjects took narrower steps when walking with higher loads (and thus at higher levels of body weight support).

IV. DISCUSSION

The decrease in step width with increasing load implies that the combination of additional inertia and body weight support had a stabilising effect on the body, reducing the required level of active lateral stabilisation via foot placement. The lack of influence of walking speed on step width is consistent with other work which found no correlation between speed and lateral measures of stability including step width [29].

The additional mass, compensated for using the body weight support system, had two effects: one of additional inertia, and also a spring-like (stiffness) effect from the lateral forces induced in the cable of the body weight support system. Results from other studies using external lateral stabilisation

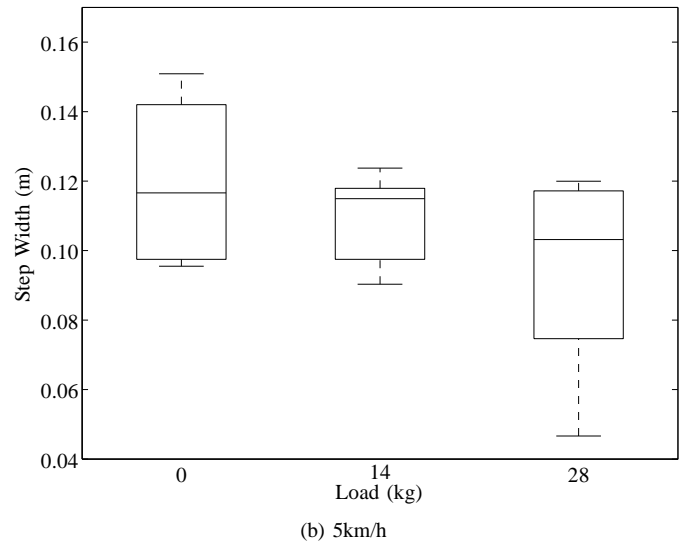
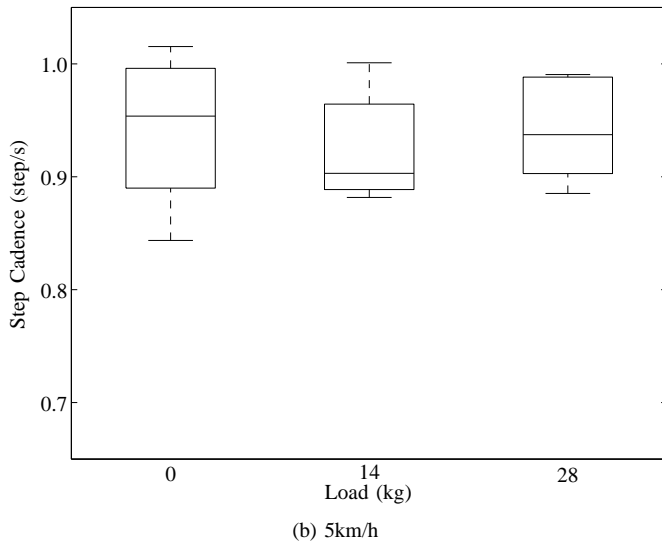
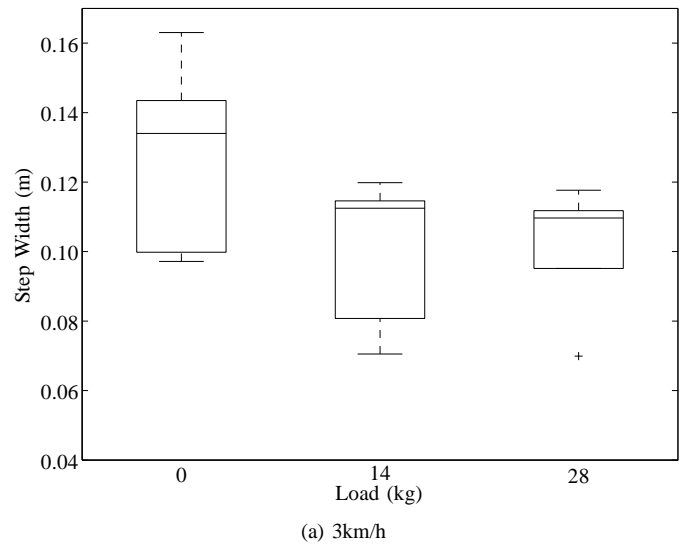
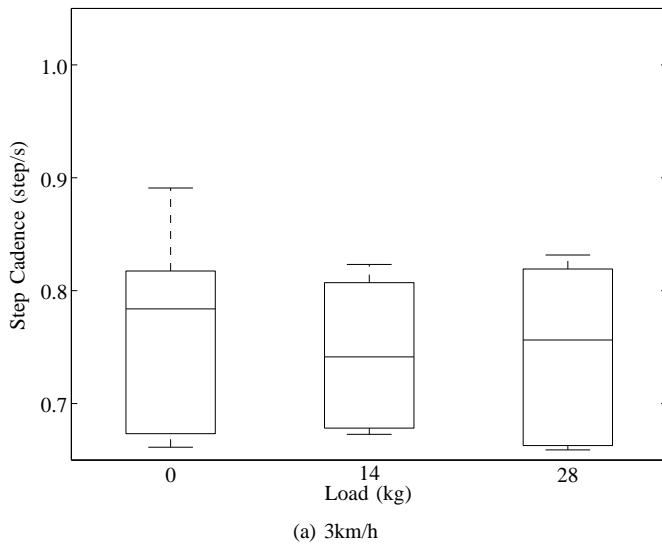


Fig. 5. Step cadences at different loading and speed conditions.

Fig. 6. Step widths at different loading and speed conditions.

via a stiffness approach have also demonstrated a reduction in step width [24], [25], although using much greater levels of stiffness than in this study.

Inertial forces would be expected to change with the (squared) frequency of the lateral movement, and yet different cadences of walking did not influence the resulting step widths taken by the subjects. This implies that the effect was mostly due to the lateral stabilisation from the body weight support system.

Models of balance using an inverted pendulum have been proposed, and have formed the basis of conditions determining necessary step widths for stability, using a prescribed initial lateral velocity as an input [30]. Models using a constant initial velocity would predict a greater required step width for a larger inertia due to the greater initial angular momentum. However, this pre-step velocity is likely to itself depend on the system's inertia. Therefore, more complex models would be required

to study the influence of inertia on step width. It may be necessary to include the kinematics of the pelvis and coupling between sagittal plane propulsion and frontal plane balance. Furthermore, energy expenditure is an important factor in the control of step width during walking [31], and is likely to be required in mathematical modelling of lateral balance.

Measurement of the lateral support force, not available in this study, could be used to accurately determine the relative contributions of body weight support and additional inertia on balance control. Alternatively, the effect of inertia alone could be investigated using a system able to maintain the body weight support cable at a vertical orientation, preventing lateral forces from being developed at all. This would require a control system of relatively high bandwidth.

Robotic gait therapy will typically have both additional inertia and stabilisation from the body weight support system. In actual clinical application, the stabilising effect could be

much greater than in this study since the magnitude of the moment from the person's weight is also reduced. Moreover, a substantial portion of a person's body weight can be needed to allow patients with a high level of impairment to perform stepping in the devices. Stabilisation effects are also seen in the sagittal plane, where patients have been observed to use the body weight support system and harness for support in anterior tilt (leaning forwards). While such additional stabilisation may be beneficial in the early stages of rehabilitation where a large degree of assistance is required, in later stages such support is likely to reduce the amount of effort required of a subject in postural and balance tasks in both frontal and sagittal planes.

V. CONCLUSION

The combination of additional inertia and the body weight support system significantly decreased the step widths used by the subjects to maintain balance in the frontal plane. The reduction in step width arose mainly due to the stabilising lateral forces developed in the body weight support cable. Since this could significantly reduce the challenge of maintaining lateral balance during walking, body weight support mechanisms in which horizontal force components are prevented may be useful in robot-driven gait training incorporating balance and postural control.

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