Lateral balance control for robotic gait training.

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***Abstract*— For the rehabilitation of neurological patients robot-aided gait training is increasingly being used. Lack of balance training in these robotic gait trainers might contribute to the fact that they do not live up to the expectations. Therefore, in this study we developed and evaluated an algorithm to support lateral balance during walking, through controlling pelvis motions. This algorithm assists the pelvis, according to a natural pelvic sway pattern, rather than attracting it to the middle of the treadmill. The support algorithm was tested on six healthy young subjects who walked on a treadmill, while different support gains were introduced. Using a higher support gain resulted in a closer approximation of the pelvic sway towards the reference pattern. Step width and step width variability reduced when the external stabilization was provided, and the stability margin increased. This indicates that artificial stabilization reduces the need for active lateral balance control. The presented algorithm to support lateral balance provides a way to assist balance in a more physiological way, compared to attracting the subject to the centre of the treadmill. Here the user is attracted/assisted towards a more natural weight shift pattern. This also facilitates a more natural input of the load receptors, which are largely involved in the regulation of muscle activation patterns and the transitions between the different gait phases.**

Keywords: gait; robotic gait training; balance training; stability; support of subtasks; assist as needed; physical guidance; extrapolated centre of mass

I. Introduction

Robot-aided gait training is increasingly being used in neurological rehabilitation. Rehabilitation robots can be used to provide more frequent, and more intensive training sessions, which are considered crucial for regaining functional mobility [1-4]. Meanwhile, they can reduce the workload of the therapist, compared to more conventional forms of manual-assisted (and body weight supported) gait training.

Despite these potential benefits, robotic gait trainers have not yet demonstrated clear advantages over conventional gait training approaches, in terms of functional outcome [5-8]. Lack of balance training in robotic gait trainers might contribute to the fact that the robots did not live up to the expectations. Active balance control is not yet implemented in most commercially available gait trainers, like the Lokomat (Hocoma AG, Switzerland), the ReoAmbulator (Motorika, USA), and the Gait Trainer (Reha-Stim, Germany). In these gait trainers the pelvis motions are constrained to the sagittal plane, whereas balance is known to be passively stable in sagittal plane but requires significant active control in the frontal plane [9]. Studies on balance control also revealed that there is little correlation between static and dynamic balance or between standing and walking balance [10, 11]. Therefore it is imperative that rehabilitation strategies also include balance training during walking.

Even in manual-assisted treadmill training of severely affected patients, balance training is limited. These patients often require a substantial amount of body weight support, which is provided via an overhead harness. This kind of setup not only provides a force in the pure vertical direction, but also in the horizontal plane. This force effectively stabilizes the body. Therefore, it is important to study how balance assistance affects the learning of walking balance. With this knowledge, more effective training methodologies can be designed for gait rehabilitation.

Walking balance is often affected in stroke or SCI patient, but is also a general problem in elderly people. The fact that these balance disorders are more dominant in the frontal plane, than in the sagittal plane, also affirms that lateral balance requires more active control. This is also confirmed by their gait parameters. To compensate for lateral instability, elderly often increase their step width, whereas step length is less affected [12-14].

Modeling, and experimental work, suggest that lateral balance control requires more active involvement than sagittal balance [9]. The active adjustments during lateral balance are also reflected in the step width, step width variability and energetic cost. Bauby *et al*. reported that lateral variability was 79% larger than sagittal variability for young healthy subjects [9]. Additionally, Donelan *et al*. showed that both mechanical work and metabolic energy increase sharply with step width [15]. Patients or elderly with imperfections in their lateral balance control are expected to require more step-to-step correction to maintain their balance. In fact: step width variability, and not step length or step time variability, discriminates gait of healthy young and older adults during treadmill locomotion [12, 13, 16]. Dean *et al*. reported similar age related effects, but only when the step width was prescribed [14].

To study how balance assistance affects learning of walking balance, several studies applied external lateral stabilizations. Donelan *et al.* provided young healthy subjects with external lateral stabilization, using elastic cords attached to the subjects' waist. This resulted in a decrease in step width, as well as step width variability and metabolic cost [17], thereby showing that artificial stabilization reduces the need for active lateral balance control. Dean *et al*. performed a similar experiment, also including elderly and reported similar effects, except that they did not observe a decrease in the step width variability during their "preferred step width" condition [14].

Domingo *et al.* used a similar setup to investigate external lateral stabilizations in a short-term walking experiment, where subjects were trained to walk on a small and wide beam. They concluded that providing lateral balance assistance can inhibit motor learning during relatively simple tasks (wide beam walking) [18]. This is consistent with the concept of "assist as needed", i.e. assistance should only be given as much as is needed to complete the task successfully.

The use of this kind of spring constructions presents a simple way to provide balance support during treadmill training of neurological patients. However there are some limitations that need to be addressed before extrapolating this kind of training towards the clinic.

First, this kind of setup limits normal arm swing. Although arm swing is not a requirement for walking balance, it reduces metabolic cost [19, 20], increases perturbation resistance [21], and is suggested to enhance gait stability during steady-state gait [22]. Thus, to provide balance training that mimics over-ground walking, normal arm swing is preferred. Second, the elastic cords also provide balance in the sagittal plane. Although the used cords are relatively long (3 m each), they also produce a restoring force in the anterior-posterior direction, which might affect gait characteristics. It also implies that not only the lateral balance is supported, but also sagittal balance, which is often not affected. Third, the stiffness of the elastic elements are subjectively determined during preliminary tests, and provide the same amount of supportive forces throughout the session, regardless of the capabilities or fatigue of the patient. Fourth, the external stabilizations might make the patient reliant on the external support, which would limit the transfer of improved balance performance on the treadmill to over-ground walking.

Finally, and most importantly, due to the setup, the patient is always attracted to the middle of the treadmill, even when the patient has a physiological pelvis motion. This limits the possibility to make movement errors, and explore the limits of their stability, which are essential for motor learning processes [23]. Similarly, Domingo *et al.* showed that the group that had greater pelvis movement variability during training, showed greater improvements in performance after training [18].

In this study we developed a lateral balance support setup that 1) allows normal arm swing, 2) decouples the lateral and sagittal balance support, and 3) allows online modification of the equilibrium position and stiffness of the corrective spring. By changing the equilibrium position of the spring the subjects only experience support when their pelvis motion deviates from a certain reference pattern, instead of a deviation from the center of the treadmill. By changing the stiffness of the corrective spring, the amount of support can be adapted to the subject's individual needs.

The purpose of this study was to develop and evaluate an algorithm that supports lateral balance control during walking, through controlling pelvis motion. Since some couplings between lateral and sagittal foot placements are reported for walking balance [9], the effect of the controller on step width and step length was measured. The support algorithm was tested on six healthy young subjects who walked on a treadmill, while different spring stiffness parameters and amplitudes of the reference motion were introduced. We hypothesized that with this controller we can selectively influence balance parameters in the lateral direction.

II. Experimental Setup and Methodology

A. Subjects

Six healthy subjects (6 males, age: 28.8 ± 3.9 years, height: 1.84 ± 0.06 m, weight: 83.5 ± 17.8 kg) participated in this experiment. All subjects gave written informed consent prior to participation.

B. Experimental apparatus and recordings

To test the balance support algorithm, an admittance controlled servomotor (C40 actuator, Moog, Nieuw Vennep, the Netherlands) was used, from now on referenced to as pelvis actuator. The pelvis actuator is connected to the pelvis via two perpendicular rods and a waist strap (Fig. 1). This kind of setup allows normal arm swing. Additionally, the waist strap contains a spherical gimbal, that allows rotation about the pelvis centre of mass, about all three axes, and ensures that only lateral forces are applied. The setup is capable of rendering a virtual mass-spring-damper system. The stiffness, damping, and the equilibrium position of the spring were controlled during the experiments. The virtual mass was set to 2 kg. The pelvis actuator is fitted with encoders that are used to calculate the position of the pelvis.

Fig. 1: Experimental setup, used to support lateral balance during walking, It comprises an admittance controlled servomotor, connected to the pelvis via 2 perpendicular rods and a spherical gimbal.

An instrumented split-belt treadmill (Y-mill, ForceLink, Culemborg, The Netherlands) was used to measure the ground reaction forces and torques below each belt. These were used to calculate the Centre of Pressure (CoP) movements during gait. The vertical ground reaction forces are also used to detect the different gait phases. Matlab Simulink (Mathworks, Natick, Mass., USA) was used to control the pelvis actuator and the treadmill at 100Hz. The pelvis position, the gait phases, and the CoP, are stored for later processing.

C. *Pelvis control strategy*

In this study we provide support based on the deviation from a healthy reference pattern. In comparison to previous studies the virtual spring of the pelvis actuator has a dynamic equilibrium position. In other words: if the subject's pelvis moves according to a certain reference, no assistance is provided. A well-known condition for standing stability is that the vertical projection of the Centre of Mass (CoM) should be within the base of support. However in dynamic situations Hof *et al*. showed that the extrapolated Centre of Mass ($XCoM$) provides a better measure to assess whether stability is maintained [24], since it accounts for the effect of the CoM velocity. The $XCoM$ is defined as follows:

$$XCoM = CoM + \frac{\dot{CoM}}{\omega} \qquad \omega = \sqrt{\frac{g}{l}} \tag{0.1}$$

with CoM the Centre of Mass, $\dot{CoM}$ its velocity, g the acceleration of gravity, and l the height of the CoM . According to this theory lateral stability is maintained when the $XCoM$ remains within the boundaries of the base of support. Therefore we defined the reference pattern of the pelvis in terms of $XCoM$, rather than CoM . This leads to the following support strategy:

$$\begin{aligned} F_{support} &= K(XCoM_{ref} - XCoM) \\ F_{support} &= K((CoM_{ref} + \frac{\dot{CoM}_{ref}}{\omega}) - (CoM + \frac{\dot{CoM}}{\omega})) \\ F_{support} &= K \cdot \Delta CoM + \frac{K}{\omega} \cdot \Delta \dot{CoM} \end{aligned} \tag{0.2}$$

where CoM_{ref} denotes the reference trajectory of the CoM , K the support stiffness, and Δ the deviation from the reference. This implies that the reference trajectory is still defined in terms of CoM , but an additional damping of K/ω is introduced.

D. *Pelvis reference trajectory*

In the current setup we are not capable of estimating, and controlling, the exact position of the CoM online. Therefore we assume the position of the pelvis and the CoM to be equal. This also means that the reference movement pattern, (the equilibrium position of the spring) needs to be defined for the pelvis. To create these reference patterns, pelvis motions from 15 healthy elderly subjects were recorded while walking at various speeds on a treadmill. During these tests the subjects were not connected to the pelvis actuator. The recorded movements of the pelvis were parameterized by defining different key events (minima, maxima etc.), which were extracted from the subjects' mean patterns at each speed. Next, the walking speed and body-height dependency of the parameters were determined by regression models. These regression models were used in this experiment to estimate the different key events for a specific subject height and walking speed. A piece-wise quintic spline is fitted between the predicted key events to create the required pelvis reference pattern. This method is described in more detail in [25].

E. *Synchronisation*

In this study we intend to selectively support balance, without affecting other spatiotemporal parameters, like cycle time. Therefore the subjects should stay in control of their cadence. To do so, the reference trajectory for the pelvis is not replayed as a function of time but as a function of the gait phase. The gait phase is based on the online estimation of the cycle time, using the vertical forces of the left and right belt of the treadmill.

F. *Experimental protocol*

Before positioning the subject on the treadmill, the body-height was measured in order to construct the appropriate reference trajectory for the pelvis. The first test consisted of a validation of the reconstructed reference trajectories. Here, the subjects walked on the treadmill at six different speeds: 0.5, 1, 2, 3, 4 and 5 kph. After a general familiarization period of 3 minutes, the subjects walked for 1 more minute at each selected speed. During this trial the spring stiffness was set to zero, and the subjects did not receive any specific instructions about how to walk on the treadmill.

During a second trial, different support gains (i.e. spring stiffness values) were tested. In this study only healthy subjects were included. Therefore, they were expected to walk according to the reference pelvis movement and, consequently, do not receive any supportive forces. In patients, the amplitude of the pelvis movement (lateral sway) is often enlarged. Consequently, balance support will be focussed on reducing their sway. To simulate this process, the amplitude of the reference pattern for the healthy subjects was reduced by 50% (reference pattern gain=0.5, see Table 1). Additionally, we also doubled the amplitude to investigate if the lateral sway increased.

The second trial was performed at a constant walking speed of 3 kph. Every condition was tested for 60 seconds. Between every condition a 30 second period of zero spring stiffness was introduced to wash out any motor learning effects/gait adaptations. The different walking conditions are listed in Table 1. All conditions were randomised between subjects.

TABLE 1: LIST OF TESTED CONDITIONS

Condition	**Spring stiffness (N/m)**	**Reference pattern gain**
1	0	-
2	1500	1
3	4500	1
4	1500	0.5
5	4500	0.5
6	1500	2
7	4500	2

G. Data analysis.

All signal processing was done with custom-written software in Matlab (Natick, Mass., USA). Of all the recorded conditions only the last 30 seconds was used for data processing. Average steps were calculated by splitting the data into individual gait cycles, based on the heel-contact events obtained from the measured vertical ground reaction force. Next, the different data blocks were normalized as a percentage of the gait cycle and averaged.

The validation of the reconstructed pelvis reference patterns was performed based on the root mean square error (RMSE) between the mean pelvis trajectory of the subjects and the reconstructed reference pattern. To compensate for lateral drift on the treadmill, the mean value of the normalized trajectories was subtracted. The RMSE was averaged across subjects for each walking speed. To quantify the timing of the reference pattern the correlation coefficient between the mean pelvis trajectory and the reconstructed reference pattern was calculated.

To evaluate if we can gradually influence the pelvis movement, by selecting a different spring stiffness, the lateral sway parameter was used. The range of the lateral sway of the pelvis, and its variability, were calculated for all subjects and conditions.

To investigate the relation between lateral sway and taking wider steps, both measures were normalized with respect to their nominal values. The nominal values were recorded during the condition in which the subjects walked with zero spring stiffness. To test if the effect of the controller is limited to gait parameters in the lateral direction, the normalized step length was also calculated. Both step length and step width are calculated from the lateral and anterior-posterior movement of the CoP.

As a measure of the lateral stability during the different conditions we used the stability margin (b) as defined by Hof *et al.* [24]:

$$b = |CoP - XCoM| \tag{0.3}$$

where b is calculated as the shortest (perpendicular) distance between the position of the $XCoM$ and the position of the ankle. The position of the ankle is calculated by taking the average position of the CoP during the single stance phase.

III. Results

A. Pelvis reference trajectory reconstruction

With the obtained regression models a set of reference trajectories for the pelvis was reconstructed for every subject (at each walking speed). For the higher walking speeds the reconstructed trajectories matched the measured data well (Fig. 2). At lower walking speeds the measured sway exceeded the sway of the reference pattern. The quality of the fit, and the timing of the trajectory, are also reflected in the RMSE and correlation coefficients (Fig. 3). Generally, the RMSE decreased at higher walking speeds and the correlation increased, indicating a better timing between the replayed reference trajectory and the actual pelvis movement.

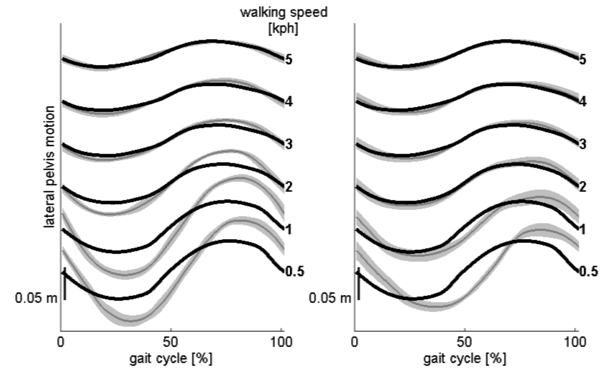

Fig. 2: Two typical examples of the actual (gray) and estimated reference trajectory (black) for the pelvis movement. The shaded area denotes the standard deviation.

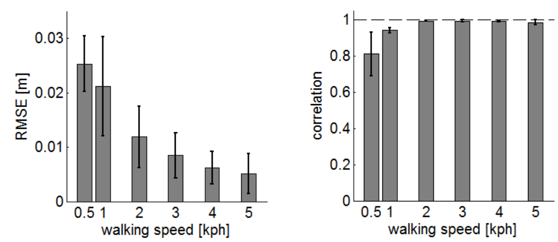

Fig. 3: Left: RMSE between actual and estimated reference trajectory for the pelvis movement. The RMSE was averaged across subjects for each walking speed. The error bars indicate the standard deviation. Right: Mean correlation coefficients between actual and estimated reference trajectories.

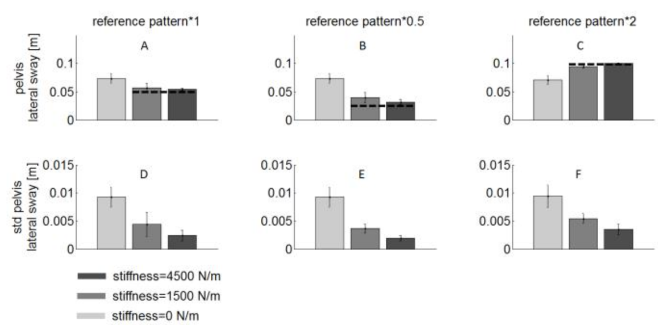

Fig. 4: Lateral sway (A-C), and standard deviation (variability) (D-F) of the sway for the different conditions. Values are averaged across subjects. The error bars indicate the standard deviation. The black dotted line represents the lateral sway of the reference trajectory

B. Lateral pelvis movement

To evaluate the effectiveness of the balance support algorithm, the lateral sway parameter was evaluated. Using a higher stiffness resulted in a closer approximation of actual sway towards the sway of the reference trajectory (Fig. 4A), when the default reference trajectory was used (reference pattern gain=1). The same behaviour was observed when the reference pattern was reduced by a factor 2 (see Fig. 4B) or multiplied by a factor 2 (see Fig. 4C). There was a clear relation between the variability in pelvic sway and the stiffness. Increasing the stiffness resulted in a reduction of the sway variability for all conditions (Fig. 4D-F). This figure also shows that most subjects had more natural sway at 3 kph than the default reference trajectory (Fig. 4A, stiffness=0)

C. Normative gait parameters

The normative gait parameters showed that the effect of the support algorithm was mainly restricted to gait parameters in the frontal plane. Increasing and decreasing the pelvic sway clearly influenced the step width, although the relative change in sway was larger than the relative change in step width (Fig. 5). The step length, and consequently step time and cadence, were not affected by a decreased sway, and were only marginally affected when the sway increased (Fig. 5).

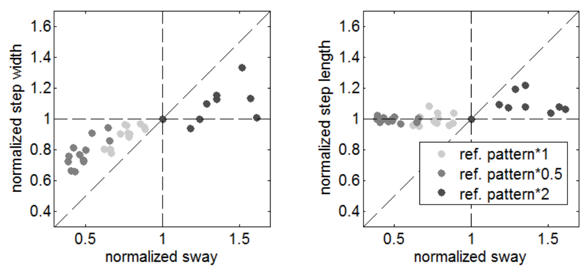

Fig. 5: Normalized sway versus normalized step width (left) and normalized sway versus normalized step length (right).

D. Stability margin

As mentioned in the introduction, step width variability and active control of lateral balance are closely related. The same applies to the stability margin, which is a criterium for maintaining balance. Fig. 6 shows both measures for the different conditions. When the reference pattern is reduced by 50%, walking becomes more stable. This is reflected in a decrease of the variability in the step width (Fig. 6B), and an increase of the stability margin (Fig. 6E). Reversely, when the reference pattern is doubled, the variability in the step width increases (Fig. 6C) and the stability margin decreases (Fig. 6F). These effects increase with stiffness. For the stiff controller the stability margin even becomes negative for some subjects. This figure also illustrates that step width and step width variability are related. A reduction in step width (Fig. 6H) is accompanied by a decrease in step width variability (Fig. 6B) and reversely (Fig. 6I and Fig. 6C).

IV. Discussion

The purpose of this study was to develop and evaluate an algorithm that supports lateral balance control during walking, through controlling pelvis motion.

For this purpose we first derived reference patterns for the pelvis. These reference patterns serve as the equilibrium position of the spring, and were based on data from 15 healthy elderly subjects. At higher walking speeds the pelvis motions of the 6 healthy young subjects matched the reference motions

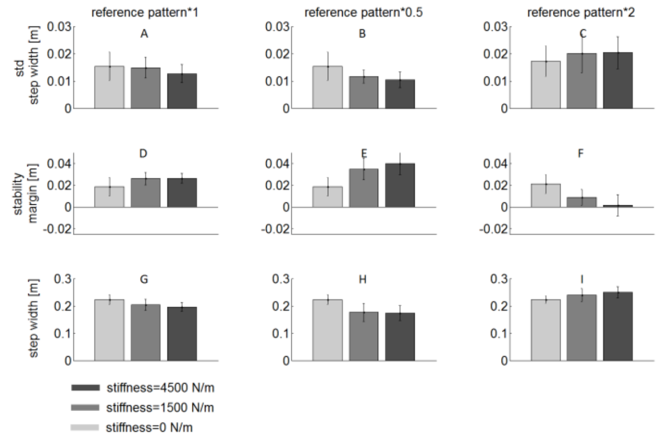

Fig. 6: Standard deviation (variability) of the step width (A-C), stability margin (D-E) and step width (G-I) for the different conditions. Values are averaged across subjects. The error bars indicate the standard deviation.

well. At lower speeds the young subjects showed more lateral sway. This might be related to the experienced inertia of the device (2 kg). However, a study on the effects of inertia, showed that adding 5.3 kg of inertia to the pelvis in medial lateral direction had a negligible effect on gait kinematics [26]. The difference in age, or the large variation of lateral sway patterns between subjects, at lower speeds, might explain these differences.

The balance controller itself was evaluated using different stiffness values and reference pattern gains. For all conditions, a larger stiffness resulted in a closer approximation of the reference trajectory, and a reduction of the variability in the sway (Fig. 4). The smallest stiffness used in our setup (1500 N/m) is similar to previous studies (1200 N/m [14], 1700 N/m [17]). In this study, the low spring stiffness was found to be effective in attracting the pelvis to the reference pattern. For this stiffness, and a reference pattern gain of 1, the maximum corrective forces were around 3 N. The largest forces were applied during the condition in which the reference pattern was doubled, in combination with the stiffest spring (4500 N/m), and reached up to around 30 N.

The effects of the balance controller were mainly restricted to the frontal plane. The step width was clearly influenced by the pelvic sway. However, the relative change in sway was larger than the relative change in step width. The pelvic sway had only little effect on the step length. Other studies on external balance support used a metronome to fix the cadence, limiting the changes in step length [14, 17].

The relation between pelvic sway and step width is also reflected in the stability margin. When the pelvic sway (and its velocity) increases more, compared to the increase in step width, the stability margin decreases. The average stability margin during the condition where the subjects walked with zero spring stiffness was around 2 cm, which is similar to the reported 2.5 cm by Hof *et al.* [24]. Increasing the lateral sway, when the reference pattern was doubled, resulted in a very small stability margin. This suggests that enlarging the pelvic sway caused gait instability, and should be considered as a perturbation rather than support. However, due to the dynamic equilibrium position of the spring, the controller will, at some moment, push the pelvis back medially, and prevent unstable situations.

The results from the tests with the different reference pattern gains also showed that the evoked changes in step width were correlated with the step width variability. Generally, providing external stabilization resulted in a decrease in step width and a corresponding decrease in its variability, which is a common finding amongst others [14, 17]. Here we also destabilized the subject, resulting in an increase in step width and step width variability (Fig. 6).

I. Conclusion

The presented balance-support-strategy provides a way to integrate dynamic balance retraining in robotic gait trainers. This will assist balance control in a more physiological way, compared to attracting, or constraining, the subject to the centre of the treadmill. Here the user is attracted/assisted towards a more natural weight shift pattern. This also facilitates a more natural input of the load receptors, which are largely involved in the regulation of muscle activation patterns and the transitions between the different gait phases [27]**.**

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