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Effect of added inertia on the pelvis on gait

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Abstract—**Gait-training robots must display a low inertia in order to allow normal-looking walking. We studied the effect of inertia added to the pelvis during walking. We attached subjects to a mechanism that displays inertia to the pelvis in the anterior/posterior (AP) direction and the lateral direction independently. During walking we measured EMG, metabolic rate and kinematics of nine subjects. We found that inertias up to 5.3 kg added in lateral direction had no significant effect on gait. We found that 4.3 kg added in the AP direction had a significant but not relevant effect on the range of motion (RoM) of pelvis AP displacement and acceleration, and on hip flexion. 10.3 kg caused a significant and relevant difference in pelvis acceleration RoM. 6 kg is estimated as the maximum inertia that gait-training robots can add to the pelvis, without affecting the gait.**

Keywords- gait training; robot; inertia; gait kinematics; metabolic rate; pelvis

I. Introduction

Robot-aided gait training is an emerging field in robotics. Several robotic gait trainers have been developed in the last two decades [1]. Studies have shown that active participation of the patient has a positive effect on the rehabilitation process [2]. To facilitate active participation, the gait trainers must provide assistance only when needed [3-5]. To implement Assist As Needed (AAN) control, the robot must be capable of following the patient's movements with minimal interaction force, when the patient does not require assistance, also known as "zero impedance control" [3] or "transparent mode."

The target of zero impedance control is to minimize interaction force between robot and subject. The remaining impedance can be expressed in mechanical impedances such as inertia, damping, friction, stiffness, and combinations. Most impedances can be compensated for completely with control algorithms. Inertia, however usually cannot be compensated for completely. In robotic gait rehabilitation, this means that the inertia of the robot is perceived by the patient. Therefore it is important to know the effect of added inertia on gait.

Different studies have shown the effects of added inertia during walking on energy consumption, muscle activity and gait parameters. In most studies 25 - 50 percent of the body mass was added. Results are an increase of energetics [6] and muscle activity [7]. Gait parameters remained unchanged [8] or change hardly (<3% [7]). The effect of pure inertia on gait kinematics has not been assessed, however the effect of added weight has. Effects of gravity have a significant effect on the gait [6], therefore the found effects caused by added weight are likely to differ from effects caused by pure inertia. Table I TABLE I. summarizes the found effects of added inertia in previous studies.

TABLE I. Effect of added inertia (+25% of body mass) to the trunk during walking

Quantity	Effect
Metabolic rate	+18% [6]
Muscle activity	+21% Soleus [7]
Gait parameters	~0 [8] – 3% [7]
Gait kinematics	unknown

To design robots for gait training, it is important to assess a threshold for inertia below which inertia has no effect on gait parameters. When the above-mentioned studies are considered in this light, there are some limitations. First, no study assessed the effect of inertia on gait kinematics. Second, all studies that assess the effect of added inertia did so by adding weights to a subject and compensating for the gravity of the weight by a body weight support system. A body weight support suspended on a fixed point has an equivalent of a stabilizing effect as a spring in a horizontal plane. Furthermore Aaslund and colleagues [7] have shown that the harness itself, without applying body weight support, has an effect on gait kinematics. Third, in the different studies relatively large added inertias (~20 kg) were used, whereas interaction control algorithms are expected to be able to reduce the displayed inertia to values below 10 kg. Fourth, none of the studies decoupled the effects of lateral inertia and anterior/posterior (AP) inertia, while controllers for these directions can be tuned independently resulting in independent (and possibly different) inertia.

The goal of this study was to assess the effect of adding pure inertia at the pelvis in AP and lateral direction and to quantify a threshold for inertia below which loaded walking resembles normal walking. This is done by quantifying the effect of inertia on gait parameters, gait kinematics, energetics, and muscle activity.

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II. METHOD

A. Subjects

Nine healthy adults (seven male, two female with a mean weight 74.9 ± 9.0 kg, height 1.80 ±0.10 m and an age of 30.9 ± 10.3 years) volunteered to be participants for this experiment. All subjects signed an informed consent before the experiment.

B. Apparatus

To add pure inertia, we designed a mechanism that connects the subject to two modules with adjustable inertias through a light-weight pelvis strap. The pelvis strap contains a light-weight bar, a rigid belt, and a trapezium construction, that allows pelvis rotation in the coronal plane.

A single module of adjustable inertia consists of a horizontal bar connected with spherical joints to a stand at one end and to the pelvis strap at the other end. Dumbbell weights are mounted on the bar. A steel wires connected to the stand and the joint with the pelvis strap assure vertical fixation of the bar, allowing only rotation of the bar and module around the vertical axis of the stand. The location of the dumbbell weight on the bar determines the added inertia on the pelvis strap, according to (1) and (2).

$$M_x^{pelvis} = \zeta^2 M_A + M_{x0} \quad (1)$$

$$M_z^{pelvis} = \chi^2 M_B + M_{z0} \quad (2)$$

Where M^{pelvis} denotes the added inertia on the pelvis, M_A (15 kg) and M_B (15 kg) are the masses of the dumbbell weights A and B, M_{x0} (0.58 kg) and M_{z0} (0.41 kg) are the inertias of the construction without the dumbbell weights at the pelvis in X- and Z direction respectively.

Parameters ζ and χ are the effective inertia gearing of the dumbbell weights A and B, determined by the location of the dumbbell weight on its bar (see table II)

C. Recordings

The effects of added inertia were assessed by quantifying kinematics, muscle activity and energetics

1) Kinematics and gait parameters

Motions were measured using an optical tracking system (Vicon Oxford Metrics, Oxford, UK). Twenty two reflective markers were attached to the human body; these markers were attached on both sides of the subject. Four markers were placed on the upper extremity i.e. shoulders, trunk chest and back. At the pelvis four markers were placed. On each leg seven markers were placed i.e. toe, heel, ankle, shank, knee, and thigh. Two extra markers were placed on the apparatus, one on the stand and one on the pelvis strap. All markers were recorded at a sampling rate of 120 Hz by means of optical tracking.

TABLE II. PARAMETER VALUES FOR X AND Z LOADING

Xload conditions			Zload conditions		
	ζ	M_x^{pelvis}		χ	M_z^{pelvis}
1	0.10	0.73 kg	1	0.12	0.64 kg
2	0.50	4.33 kg	2	0.36	2.33 kg
3	0.80	10.18 kg	3	0.57	5.31 kg

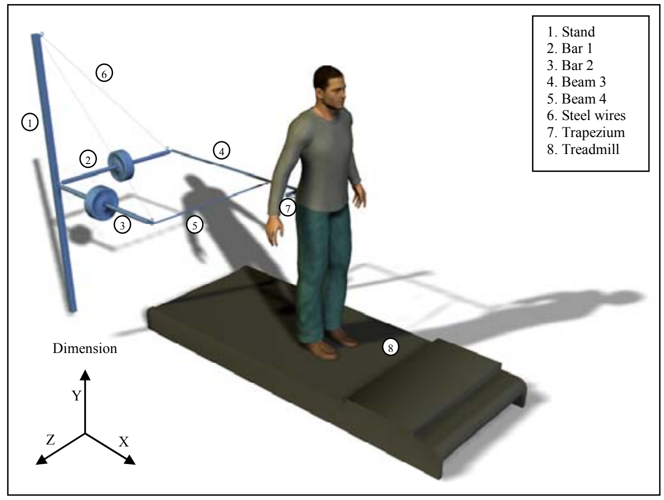

Figure 1. Experimental setup for applying inertia on the pelvis in x- and z direction.

2) Muscle activity

The muscle activity was measured by recording the electromyography (EMG) from eight different muscles of the right leg: (1) the *gluteus maximus*, (2) *gluteus medius*, (3) *rectus femoris*, (4) *vastus lateralis*, (5) *biceps femoris* (6) *gastronemius medialis* (7) *soleus*, and (8) *tibialis anterior*. The analog signals were sampled at 1024 Hz and recorded with a Bagnoli system (Delsys, Boston, USA). Amplified EMG data was synchronized with the VICON System.

3) Energetics

The energy expenditure was measured by the Oxycon Pro system (Jaeger, Hoechberg, Germany). Subjects were connected to the Oxycon with a flexible tube making an airtight seal to a facemask, measuring oxygen consumption (VO2) and volume expiration (VE). The heart rate of the subjects was measure at the index finger by a pulse-oximeter . Every five seconds (0.2 Hz) all parameters were measure and stored on the personal computer that was connected to the Oxycon.

D. Experimental protocol

The experiment started with two conditions in which subject were walking on a treadmill at 1.5 km/h and 4.5 km/h without being attached to the system, called “no load” conditions (NL). These trials were followed by a randomized sequence of the added inertia- and speed conditions. For both loading directions, three conditions were used (see table II). For speed two conditions were used: 1.5 km/h and 4.5 km/h. Combining the three parameters, resulted in 18 different loaded conditions.

The loaded conditions with minimum load in both X and Z directions are the baseline conditions (BSLN). The two baseline conditions (at two different speeds) were validated against the no load conditions.

All 20 trials consist of three-minute walking.

E. Data processing

For each trial only the last 30 seconds of data recording were used in analysis, in order to eliminate transition effects.

Motion data was converted to joint- and segment kinematics using Analyse [9]. 3D joint angles of the ankle, knee, and hip were analyzed. 3D centre of mass (CoM) of the pelvis segment were analyzed.

Data were split into strides based on left heel contacts. These were identified with the marker data [10]. EMG and kinematics data were divided into steps. Furthermore, gait parameters i.e. cycle time, stance time, swing time, double support time and step width were calculated.

F. Statistical analysis

First we tested whether the NL conditions differed significantly from the BLSN condition to assess whether merely attaching the mechanical setup already affected the walking pattern. Subsequently we assessed the effects of the different loads.

To asses whether inertia had a significant effect on gait, we performed a three-way (velocity, M_x, M_z) repeated measures (ANOVA). In this paper we only regard the main effects of M_z and M_z. In all tests a significance level of $p<0.05$ was used. Significant effects were evaluated on relevance.

To assess the relevance of found significant differences we took the absolute parameter change relative to the BSLN and compared that with twice the (within-subject) standard deviation of the BSLN condition averaged over all subjects. Parameters differences that are larger than the average double standard deviation of the BSLN conditions are considered a relevant change.

III. Results

A. Baseline validation

When comparing the NL conditions with the BSLN conditions, no changes in parameters were found when walking with the device with the minimal added inertia ($p>0.05$). So, walking with the minimal inertia applied resembles free walking on a treadmill.

B. Effect of loading

1) Kinematics

a) Pelvis Centre of Mass position and – acceleration

Inertia in X direction caused a significant decrease in the range of motion (RoM) of the pelvis centre of mass (pcom) in X direction in position (see figure 2 top). Inertia in Z direction did not cause a significant change on the RoM of the pcom (see figure 2 bottom).

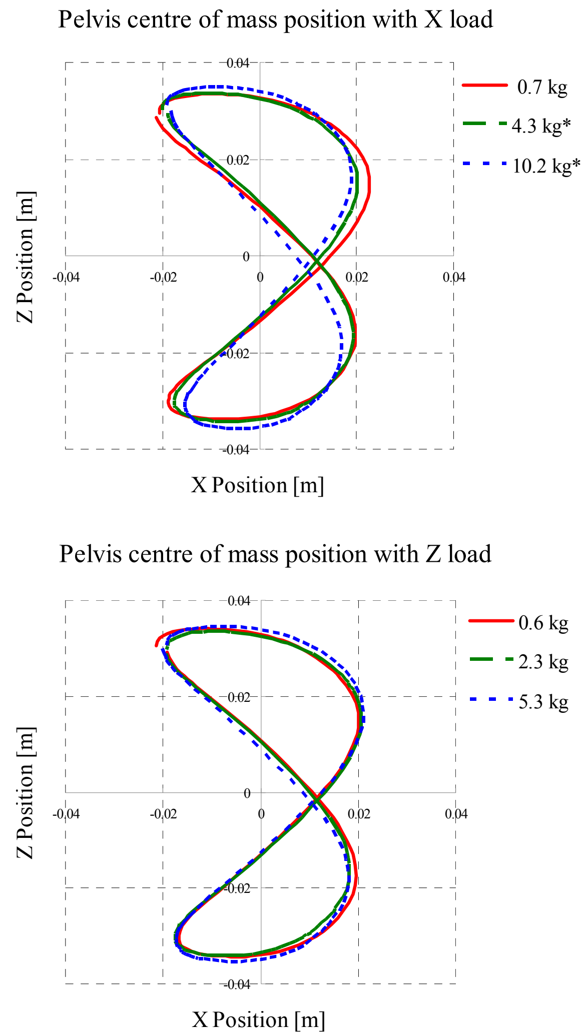

Figure 2. Effect of inertia on pelvis centre of mass. X load has a significant effect on pelvis RoM in X direction ($F(2,6)=15.042, p<0.05$)

TABLE III. EFFECT OF ADDED X INERTIA ON PELVIS ROM

		Added inertia in X direction		
		0.73 kg (BSLN)	***4.33 kg***	***10.18 kg***
Position	X [mm]	49.11 (2.14)	46.72 (1.89)*	43.00 (1.87)*
	Relative difference	-	-4.87%	-12.44%
	Absolute difference	-	-2.39	-6.11
	Z [mm]	55.03 (3.78)	55.40 (3.49)	56.51 (3.90)
	Relative difference	-	0.67%	2.68%
	Absolute difference	-	0.37	1.48
Acceleration	Pelvis RoM X [m/s^2]	3.87 (0.22)	3.55 (0.15)*	3.05 (0.12)*
	Relative difference	-	-8.15%	-21.08%
	Absolute difference	-	-0.31	-0.81†
	Pelvis RoM Z [m/s^2]	1.56 (0.10)	1.55 (0.07)	1.59 (0.09)
	Relative difference	-	-0.63%	1.95%
	Absolute difference	-	-0.01	0.03

Values are means (S.E.). Note: *significant main effect of load from BSLN, P<0.05; †Relevant effect of load ($2\bar{\sigma}_{x=0.73kg}$ = 0.52)

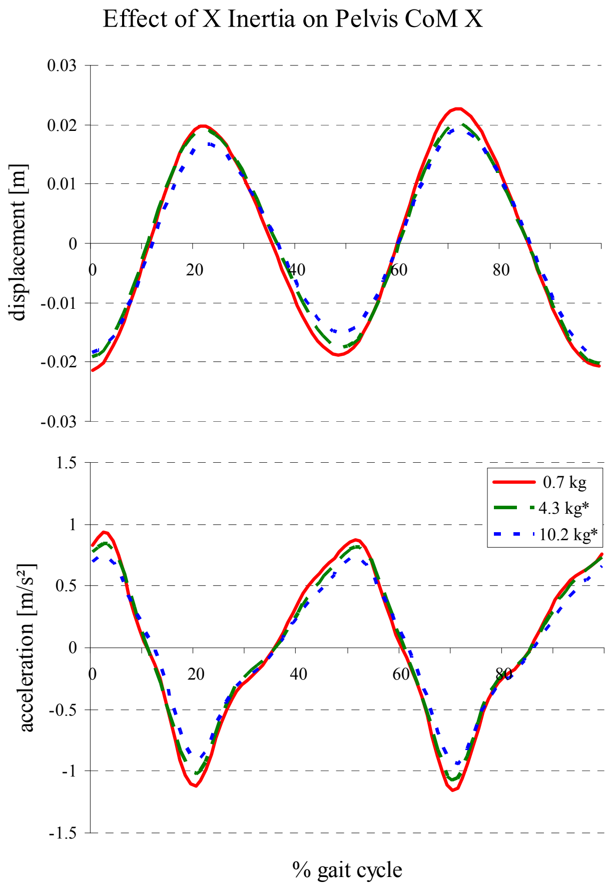

Figure 3. Effect of Inertia on Pelvis X motion, i.e. displacement (top) and acceleration (bottom)

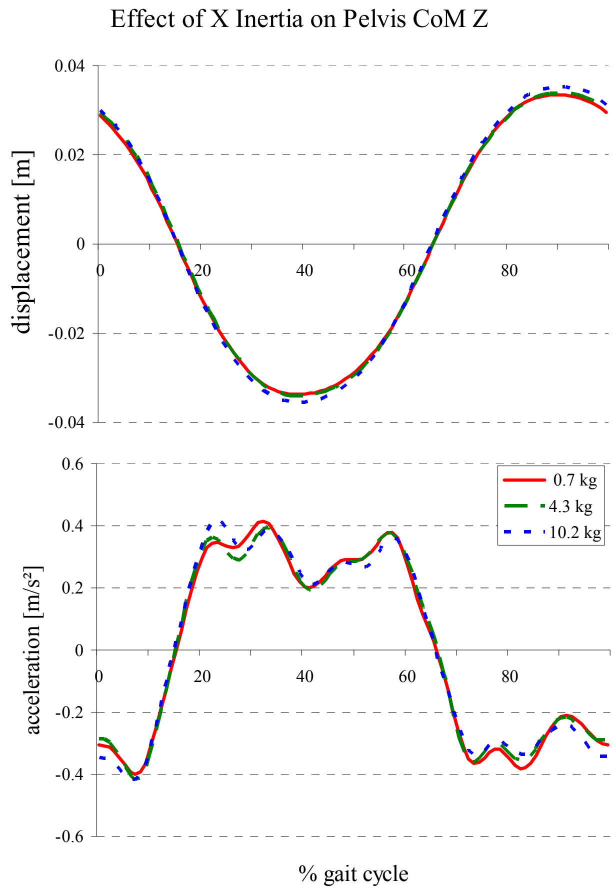

Figure 4. Effect of inertia on pelvis Z motion., i.e. displacement (top) and acceleration (bottom)

Inertia in X direction also has a significant effect on the acceleration of the pcom in X direction (see figure 3). This effect is also relevant (see table III).

The effect of inertia in X direction on acceleration in Z direction is not significant (see figure 4).

b) Joint angles

The X load had a significant effect on the hip flexion – extension RoM and hip abduction-adduction RoM. These changes were less than one degree and found not to be relevant. The other joint angles showed no significant change on either load (see table IV). Figure 5 shows the average joint trajectories of the three X loads.

2) Gait parameters, energetics and EMG

No significant differences were found in gait parameters i.e. cycle time, double stance ratio, stance time, swing time, and step width. Also on energetics (VO2, heart rate) and EMG, no significant difference was found. There were, however, significant effects of speed.

TABLE IV. EFFECT OF X LOAD ON JOINT ANGLE ROM.

	0.73 kg (BSLN)	***4.33 kg***	***10.18 kg***
Hip flexion –extension RoM [deg]	37.03 (1.05)	36.80 (0.90)*	36.12 (0.86)*
Hip abduction adduction RoM [deg]	12.23 (0.91)	11.82 (0.85)*	11.31 (0.80)*
Knee flexion – extension RoM [deg]	61.29 (1.96)	60.99 (1.73)	59.88 (1.65)

Values are means (S.E.). Note: *significant main effect of load from BSLN, P<0.05;

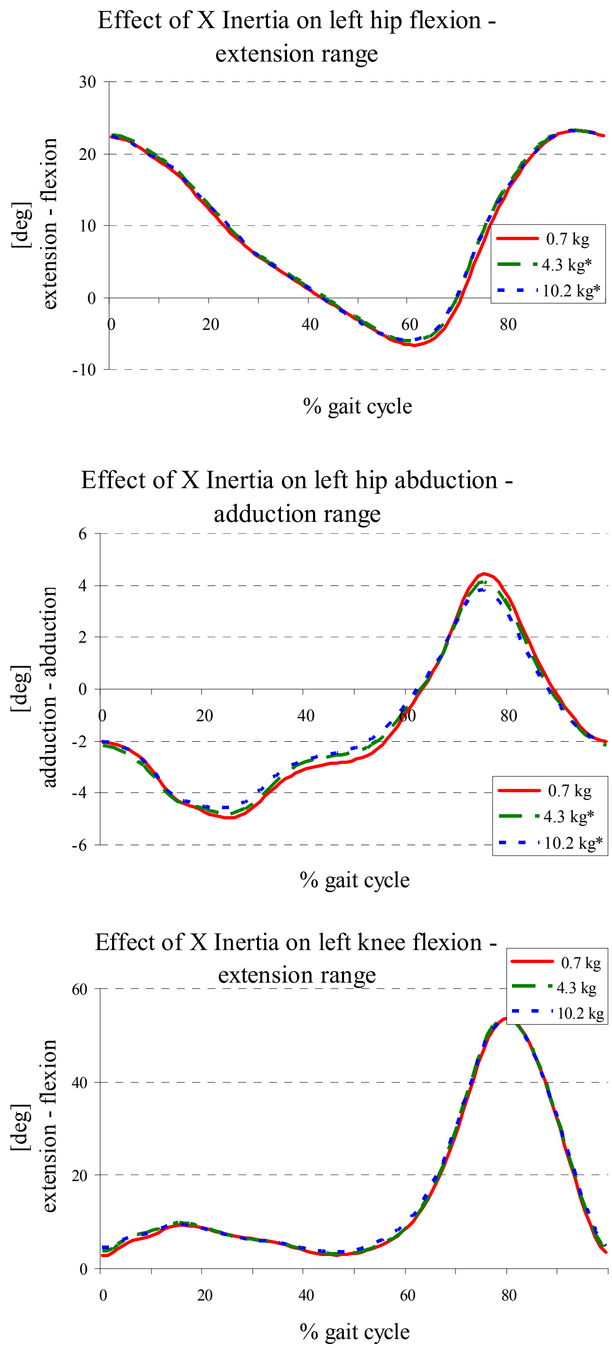

Figure 5. Effect of X load on hip flexion – extension (top), hip abduction – adduction (middle), knee flexion – extension (bottom)

IV. Discussion

The goal of this study was to assess the effect of added inertia on gait. Contrary to previous studies, we decoupled the anterior/posterior inertia and lateral inertia and did not add inertia in the vertical direction to the pelvis.

We found that inertia added on the pelvis in the AP direction has a significant effect on the RoM of the pelvis centre of mass in the AP direction. At 10.2 kg the effect was also relevant i.e. more than twice the standard deviation of the baseline condition. The decrease of position and acceleration can be accounted for by Newton's law; the inertia is connected directly to the pelvis and therefore, when force remains unchanged, the acceleration will decrease. Decrease of acceleration of a periodic movement implies a decrease of the RoM.

The fact that this phenomenon is not observed in lateral loading can be ascribed to the relatively low loading in lateral direction (half of forward/aft loading), and by the relatively low acceleration in lateral direction [11].

Although significant changes were found in hip flexion extension RoM, the difference was within one degree and consequently irrelevant.

Our results deviate from results from previous studies, where muscle activity [7] and energy consumption [6] both increased at loads of 25 percent of the body mass. In our study the highest load was 13 percent of the body mass (in AP direction), which did not result in any significant change in both muscle activity and energy consumption. The major difference between previous studies and ours is that we did not apply inertia in the vertical direction. We therefore assume that increase in muscle activity and energetics is largely attributed to the vertical inertia.

During trials subjects were asked if they felt the load (although we did not assess this in a systematic way). Several subjects mentioned they felt the presence of the load, both in lateral and AP directions. The maximum loads were sensed more often than the medium loads. This is consistent with the results from studies by Ross and Bodie [12, 13], who have found that the just noticeable difference (JND) for mass is 10 percent. On a 75 kg subject, the head-arm-trunk (HAT) segment weights about 46 kg. The JND for the HAT segment therefore is 4.6 kg. This JND is within the range of our loaded conditions, which can account for the noticeability of the loads.

V. Conclusion

This study showed a significant effect of inertia added in the AP direction on pelvis during walking. 10 kg of added inertia in the AP direction has a relevant effect on gait. The threshold below which loaded walking resembles normal walking for added inertia in AP direction is between 4 and 10 kg. For the design of gait-training robots the reflected inertia on the pelvis in the AP direction should be maximum 6 kg.

For added inertia in the lateral direction, no significant effect was found on gait. Due to the smaller accelerations in lateral direction compared to AP direction, the effect of inertia is expected to be smaller, and therefore would require a second study with larger inertias. Until then, for gait rehabilitation robots also 6 kg is recommended as maximum reflected inertia on the pelvis in lateral direction.

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