Evaluation of a Virtual Model Control for the selective support of gait functions using an exoskeleton

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***Abstract*— Robotic gait trainers are used all over the world for the rehabilitation of stroke patients, despite relatively little is known about how the robots should be controlled to achieve the optimal improvement. Most devices control complete joint trajectories and assume symmetry between both legs by either a position or an impedance control. However we believe that the control should not be on a joint level but on a subtask level (i.e. foot clearance, balance control). To this end we have chosen for virtual model control (VMC) to define a set of controllers that can assist in each of these tasks. Thus enabling the exoskeleton to offer selective support and evaluation of each substask during rehabilitation training. The bottleneck of the VMC performance is the ability to offer an end point impedance at the ankle as the arm between the joints is largest here. This endpoint impedance is evaluated in this paper to show the ability of our exoskeleton to offer the required moments to support all the gait functions defined in this paper. We have shown that it is possible to implement the VMCs necessary for selective support of gait functions using series elastic actuators with a non-linear transmission. For the vertical direction we measured an stiffness of 5kN/m for all ranges at frequencies of up to 1Hz as a near ideal spring. In the horizontal we measured op to 0.5kN/m in the same frequency range. The crosstalk between the vertical and the horizontal directions has been shown to be small. This means that it is possible to selectively offer forces in either vertical or horizontal directions.**

I. Introduction

Due to aging of the population, growing numbers of people are affected by impairments of their motor system, caused by diseases like stroke. Treatment of stroke patients is very labor intensive. Combined with the total number of people suffering from a stroke this makes rehabilitation therapy costly. Reducing the costs of therapy would make robotic aids attractive to rehabilitation centres. 'Robotic therapist's' are meant to make rehabilitation more effective for patients and less demanding for therapists [8], [5], [7]. This claim is based on the assumptions that: - intensive training improves both neuromuscular function and all day living functionality [9].

- a robot is able to train a patient at least as effectively manual training [13], [12],

-a well reproducible and quantifiable training program, which is feasible in robot assisted training, would help to obtain clinical evidence and might improve training quality [13].

The most common choice for a rehabilitation trainer design is a exo-skeleton [6]. The foremost reason for this is that the robot has no unnecessary degrees of freedom. As the degrees of freedom of the exoskeleton are the same as the patient's degrees of freedom, protection for hyperextension or flexion of joints could be realised physically in the joints. This makes the device intrinsically safe for undesired movements of the exoskeleton. Currently there are robotic rehabilitation devices under development for both the upper and lower extremities [7], [3]. Our exoskeleton, LOPES, is designed for gait recovery training after stroke. It is meant to be used in treadmill training in order to keep the patient accessible to the therapist [15].

Commercially available gait rehabilitation robots generally use a form of optimal gait pattern with or without adaptive algorithms [5]. The patient is forced or guided along a set reference trajectory describing the entire gait pattern. The patient will be trained only in this pre programmed cycle which is not necessarily the optimal pattern for that patient. We claim that passive walking can not be considered task specific training, as the patient is not carrying out the task himself. Recent studies have shown indications that task specific training leads to better results in the relearning of motor functions [14], [10]. Several groups [3], [7] are developing impedance based support devices to accommodate for more active participation. However these devices still use a (symmetrical) reference pattern as a basis. LOPES aims to offer a task specific training for patients by defining different tasks within the gait cycle and supporting those tasks separately depending on the patients needs. e.g. If a patient is unable to effect sufficient foot lift the robot will support the foot lift but will not be active in the rest of the gait cycle. This will hopefully lead to more active walking from the patient's side and a more task specific training.

The proposed control method for implementing this training in a rehabilitation robot is Virtual Model Control (VMC). This method has been implemented in the control of several 2D walking robots [11] and 3D walking robot models and in human interactive robots [1]. The method maps the desired end-point forces to joint torque references that are need to be exerted by the exoskeleton to offer this end point force. The advantage of this method is the relative ease of translation from a required gait function to a control algorithm. The use of VMC in combination with the bowden cable transmission needs to be evaluated for use in lower-extremity powered exoskeletons. The advantage of using VMC is, amongst others, that a virtual interaction force can be defined at any point. However due to constraints on actuator power it is not possible to cancel out the system inertia. Play and other non-linear effects in the robot may also influence the system ability to accurately offer the desired impedance. In this paper we will quantify the ability of an exoskeleton actuated with bowden cable driven series elastic actuators to simulate impedances necessary to support impaired gait. Based on this evaluation we will identify the factors that limit the system ability to simulate the required impedances. We will be using three criteria. First the bandwidth at which a system is capable of offering the proper impedance. Second the maximum impedance that the system is capable of offering. And third the ability of the system to offer an impedance in a specific direction without crosstalk.

The goal of this article is to analyze the applicability of VMC in a gait rehabilitation device powered with bowden cable driven series elastic actuators. To this end we have evaluated the ability of our prototype to realise the impedances needed to support impaired human gait. The criteria we have used to determine the effectiveness include the bandwidth, the measured impedance versus the required impedance and the selectiveness of the support.

II. Methods

A. Virtual Therapist

Virtual model control is a motion control framework that uses simulations of virtual components to generate desired joint torques. These joint torques create the same effect that the virtual components would have created, had they existed, thereby creating the illusion that the simulated components are connected to the real robot. Using virtual components such as inertias, springs and dampers it is possible to simulate any interaction that a therapist would usually have with a patient. As an example the foot clearance will be handeled:

The VMC model consists of a spring damper system connected to the ankle with a stiffness Ky. The required force Fy is calculated from the virtual spring Ky and the deviation of the reference vertical position (y_ref) from the vertical distance (y):

$$F_y = K_y(y\_ref - y) \tag{1}$$

The required vertical force is mapped to the torques at the hip and knee joint. The upper (lU) and lower leg length (lL) and knee and hip angles are used to determine the relation between the applied moments and the required force to be exerted by the VMC. The forward kinematic map from the hip frame to the knee frame can be written as follows:

$$^h_kX = \begin{pmatrix} x \\ y \end{pmatrix} = \begin{pmatrix} \text{lUsin}(\theta_h) + \text{lLsin}(\theta_h - \theta_k) \\ -\text{lUcos}(\theta_h) - \text{lLcos}(\theta_h - \theta_k) \end{pmatrix} \tag{2}$$

where the h and k stand for hip and knee and θ is the joint angle. Differentiation to the generalized coordinates $(\theta_h \theta_k)$ we get the Jacobian:

$$^h_kJ = \begin{pmatrix} \text{lUcos}(\theta_h) + \text{lLcos}(\theta_h - \theta_k) & -\text{lLcos}(\theta_h - \theta_k) \\ \text{lUsin}(\theta_h) + \text{lLsin}(\theta_h - \theta_k) & -\text{lLsin}(\theta_h - \theta_k) \end{pmatrix} \tag{3}$$

The Jacobian relates the VMC force to the joint torques :

$$^h_k\tau = {}^h_kJ^T F_y$$

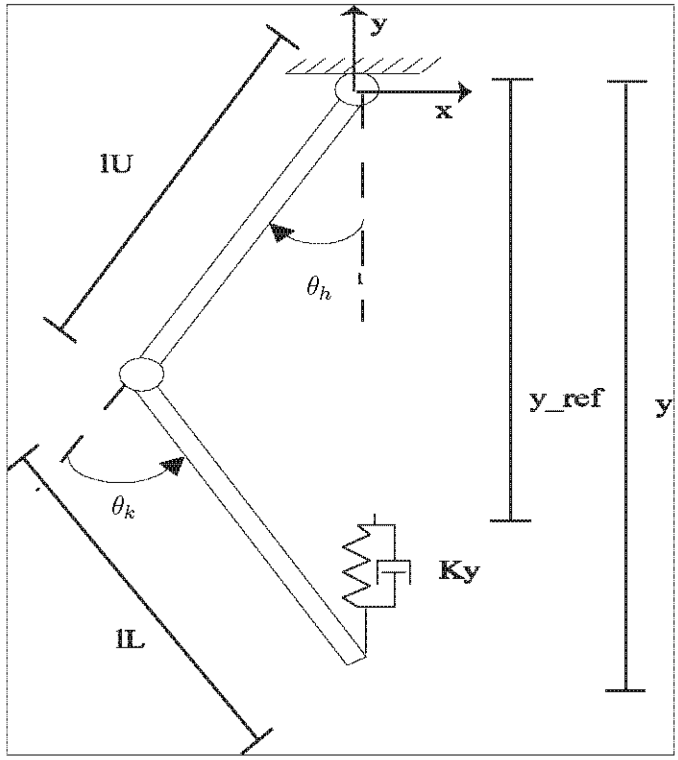

Fig. 1. Schematic representation of the VMC and the exoskeleton. Ky is the virtual spring stiffness, the vertical distance between the hip and ankle, y_ref the reference trajectory height, lU the upperleg length, lL the lower leg length and θ_k and θ_h respectively the knee and hip angles.

These joint torques references are then offered to the inner force control loops and the torques is exerted on the patient.

Following a model study [16] it was shown that several basic gait functions are enough to generate a viable gait pattern. We assume that if we have a full set of virtual models describing support for each of these subfunctions we have a sufficient support fir the entire gait cycle. We have looked at the requirements for each of these virtual models (VMs) and have defined the critical values for the necessary supporting impedance to support each gait function. Based on the criteria we have defined the extreme cases and used these as the evaluation criteria for our experiments. Wether these VMs will support each gait function will be shown in future work. At the moment we have implemented and tested one VM and this is presented in the companion paper [2]. The full list of gait functions that we assume will support the gait is:

1) Balance in the sagital / lateral plane
2) Step Width
3) Foot clearance
4) Step length / gait speed
5) Support weight Bearing

When not supporting a the robot should leave the patient in charge this means that the patient should experience unhindered walking. This is an important function during gait training of stroke patients as they quite often are only paralyzed on one side of their body.

For each subfunction we have defined a supporting

VM (figure II-A). In normal operation it would be more likely that only a few VMs would be active during a gait cycle. The therapist will be able to determine the set points and coefficients for the different VMs and thus determining the amount of support which is offered to the patient. Per VM we will also indicate settings that a therapist could use to influence gait. Finally we will define the criteria that these VMs set for our system.

Unhindered walking: While the patient walks within desired boundaries, (s)he should ideally not feel the robot.

VM Implementation: The robot needs to actively compensate for the friction of the exoskeleton and reflected mass of the motors. The inertia of the exoskeleton will not be compensated as the accelerations during stoke gait are small. We will not actively compensate for the weight of the leg parts of the exoskeleton as weight compensation would influence the swing properties of the leg. Setting: This VM has no settings.

Criteria: Minimal joint level impedance at high bandwidth.

Balance: The balance of the patients needs to be supported in the mediolateral plane.

Implementation (VM1): Lateral balance can be supported by a combination of two spring-dampers connected to the COM (VM1). Settings: In the lateral plane the therapist could set the maximum excursion of the COM with respect to the centre line. This excursion would be dependent on the current gait phase leaving the timing to the patient. In the sagital plane the therapist could set the excursion of the COM with respect to the trailing stance leg.

Criteria: Maintaining the balance of walking is the slowest function of the exo-skeleton. The forces when exerted on the pelvis remain low. We have estimated, based on our observations, that a stiffness of 1kN/m would be adequate based on observations at rehabilitation clinics.

Step Width: The step width is a determining factor for lateral stability. Wider steps lead to more stable but less efficient walking.

Implementation (VM2): A spring damper combination connected to the ankle would be adequate to simulate the therapist. Settings: The therapist would be able to set the position of foot of the swing leg with respect to the stance leg at heelstrike. Also a scaling factor for the position trajectory of the ankle during swing phase could be used for added foot clearance.

Criteria: A relatively low impedance in the sagital plane as only the swing leg needs to influenced at relatively low frequencies.

Foot Clearance: Sufficient foot clearance during swing phase.

Implementation(VM3) : This VM is implemented as a virtual roller skate. The foot can move freely in the horizontal plane but is not able to move below the minimal height ensuring the foot does not hit the floor. Settings: The therapist would be able to set the minimal ankle height during the swing phase. During stance phase this VM is inactive.

Criteria: A large stiffness damping is needed in the vertical direction is needed while in the horizontal direction the impedance should remain close to zero.

Step length / gait speed: The step length influences the sagittal balance and the gait speed. A certain step length can be beneficial to a certain patient by either catering to their inability or reducing energy cost.

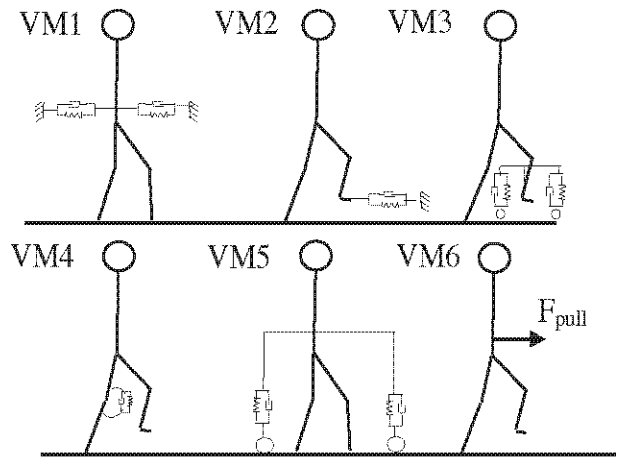

Fig. 2. Examples of Virtual Models(VM) to support gait. VM 1 supports the balance of the patient. VM2 assist the patient in the placement of the foot in the sagittal and frontal plane, which is important for dynamic balance and the speed of walking. VM3 enforces sufficient foot clearance using a virtual granny walker connected at the ankle. VM4 helps to stabilize the knee. VM5 is a virtual granny walker (partial) supporting the patient's weight. VM6 increases the patient's push off. (*is implemented)

Implementation (VM2): A spring damper combination connected to the ankle would be adequate to simulate the therapist.Settings: The therapist would be able to set the endpoint position of the foot, ie the position of the foot just before heelstrike, and the time at which this foot needs to reach this endpoint. Simultaneously VM1 needs to accelerate the COM to accommodate for the increased frequency.

Criteria: A relatively low impedance is needed in the lateral plane as only the swing leg needs to be influenced at relatively low frequencies.

Weight bearing:

Implementation (VM4 and 5):Whole body weight can be supported by implementing a virtual granny walker connected to the COM (VM 5). Implementing a virtual torsion spring on the knee just before loading stabilizes the knee (VM4)

Criteria: A large impedance is needed to support the weight of the patient. This VM needs to react quickly when a patients buckles during or following a heelstrike so a high bandwidth is needed.

In this study we will only take the VMC's working the sagittal plane in to consideration due to the limitations of our prototype. With the exception of VM4 all of the VMs used for the sagittal plane, although having different setpoints and impedance values, can be seen as a impedance between the hip and the foot of the exoskeleton. Of the given VMC's the weight support is the most demanding in the vertical plane. In order to support the weight of the patient at least 5 kN/m impedance is needed per leg. The 5 kN/m would a patient with the maximum allowed weight (90 kg) would have less than 10 cm deflection leading to a knee angle that will allow normal walking. The frequency of loading of the legs is

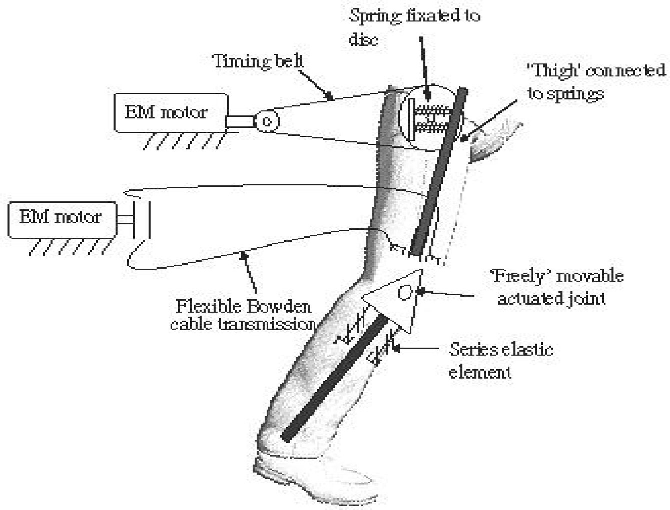

Fig. 3. A) Prototype lower extremity exoskeleton. The prototype has 2 actuated degrees of freedom: The hip actuated through a series elastic actuator B) and the knee actuated through a distributed series elastic actuator C) and [17]

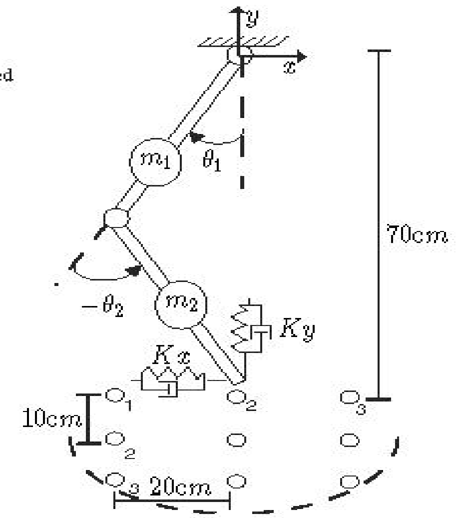

Fig. 4. Schematic representation of the prototype leg. The grid of dots show the different positions at which the impedance was measured. This grid spans the entire motion range of the ankle during CVA gait. The dashed curve shows the maximum position of the exoskeleton ankle.

roughly 1 Hz. However in order to support buckling knees it is important for the system to be as fast as possible. Thus fo the vertical direction we have want a mechanical stiffness of no less than 5 kN/m per leg with a frequency of no less than 2 Hz. For the horizontal direction the foot placement is the most demanding. Based on therapist data [4] and an assumption on the weight of the leg we assume that a stiffness of 0.5 kN/m is necessary. The frequency of this movement is roughly 1 Hz and we assume that a 2Hz response should be adequate for our purposes.

B. Experimental Setup

The prototype which has been used for this study consists of a single actuated leg (fig 3. This setup can be used to test all sagittal based VM's. The actuation consists of rotary compliant actuators. These actuators allow for gentle interaction between the robot and the patient. This design also allows for the motors to be taken from the frame and thus reducing the mass that moves with the patient. The current prototype has two different types of series elastic actuator (SEA): a Bowden cable driven SEA [17] on the knee and a time belt driven rotary SEA on the hip. For the hip actuator we used a Maxon EC45 motor with a 43 to 1 reduction. On the knee we used a Berger Lahr SER 3910 motor with a transmission of 1 to 8 reduction. The serial springs (series elastic elements) had a stiffness of 90kN/m. The transfer from reference torque to measured torque of these actuators are comparable and have a bandwidth of >20 Hz for small forces (< 20Nm) and >10 Hz for large forces(< 50Nm).

C. Experiments

The aim of these experiments is to see whether the desired virtual impedance can be realised as a mechanical impedance between the ankle of the exoskeleton and the hip joint. In order to identify the impedance the system was perturbed at the ankle along the vertical and the horizontal axes (fig 4).The system was perturbed using a DC motor in series with a play free spindle. This linear actuator functioned as a position source moving the lower end of the exoskeleton shank (roughly at the position of a persons ankle) at 9 points spanning the movement of the ankle (fig 4). The disturbance was offered in two directions: pure horizontal and pure vertical. Several impedances were measured per position. The values of the impedances to be simulated are shown in table I.

We used a crest optimised multisine with a amplitude of component sine waves of <1 cm amplitude as a perturbation signal with a decreasing amplitude for higher frequencies. The frequency content of the sine wave was between 0.1 and 4 Hz . The horizontal positions are at x=-20,0 and 20 cm. The vertical positions are at y=-70,-80 and -90 cm. Both x and y are measured with respect to the hip axis.

The desired behavior of the controlled system should respond at the ankle as a spring with the set value stiffness. However due to power considerations we do not compensate for inertia and thus we will find the inertial properties of the system in the measured impedance in combination with some residual friction. The apparent mass of the structure can be written as [18] :

$$m = \frac{(m1 + m2 - m2cos(2\theta_2) + m1cos(2\theta_1 + 2\theta_2 - 2\theta_3)}{sin(\theta_2)^2} \tag{4}$$

With θ_1 the hip angle θ_2 the knee angle, m1 and m2 respectively the thigh and shank mass of the exoskeleton. The endpoint(ankle) of the system will behave as a mass, spring damper system with, mass m, K as stiffness setting for the impedance controller and the residual damping [17]. The behavior of the system will then be compared to a modeled spring, mass and damper system.

TABLE I

THE REQUIRED STIFFNESS COEFFICIENTS OF THE VM IN THE X AND Y DIRECTIONS, K_x AND K_y RESPECTIVELY. THE AMPLITUDE IS THAT OF THE PERTURBATION MULTI SINE SIGNAL.

K_x[N/m]	$K_y[N/m]$	amplitude[cm]
0	0	1
500	0	1
1000	0	1
0	1000	0.5
0	2000	1
0	2500	1
0	5000	0.5

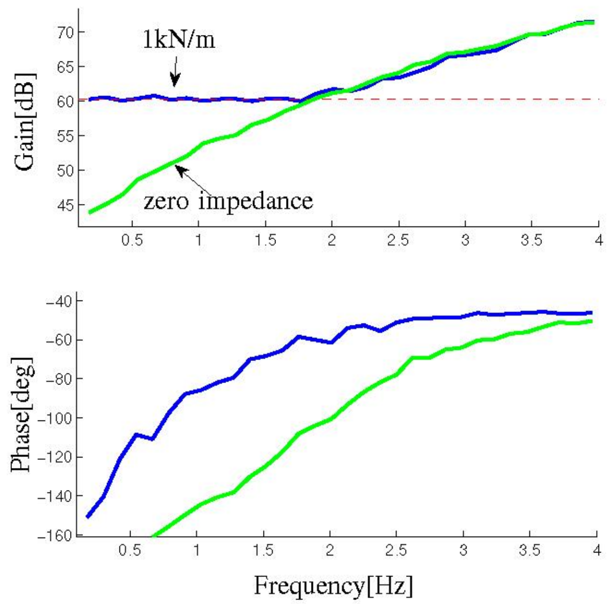

Fig. 5. Bode plot of two measured impedances at the center grid position (fig 4) (the gray line shows the zero impedance mode and the black line shows the 1kN/m results). These measurements give a typical situation for all orientations and postitions

III. RESULTS

Zero impedance

In fig 5 we see a bode plot of the measured impedance for 1 kN/m and a 0kN/m setting. The gray line shows the zero impedance line and this shows a line with a constant gradient typical to the behavior of a mass. The black line shows the 1 kN/m results and this shows a horizontal line up to 1.8 Hz meaning the system behaves as a spring damper with a 1 kN/m stiffness up to this frequency. At higher frequencies we can see that the two lines converge implying that the system can no longer simulate the desired stiffness and the mass properties become dominant. This result was a typical for all positions and desired impedances measured . With the leg in this mode we have had several people walk in the exo-skeleton and we have found that the resistance of the exo-skeleton joints did not hamper movement noticeably to the test persons. The maximum interaction force measured at the elastic elements was 1 Nm per joint during normal walking which is roughly 2% of the maximum exerted during normal walking.

Impedance measurements

A bode plot of a typical measured impedance versus a bode plot of the corresponding model is shown in fig 6. Up to 1 Hz the system functions as a ideal spring, a gain value of close to the desired stiffness and a phase of close to -180 degrees. Above this value the system inertia becomes the dominant factor in the system impedance, which is shows by the fact that at higher frequencies the lines have a constant positive gradient. However the VMC controller does realize the desired spring stiffness for frequencies up to 4Hz (maximum frequency measured). This means that for all possible voluntary motions from the patient the system will perform predictably. For the 5kN/m measurements the motors did reach peak torques making it necessary to reduce the amplitude of the perturbation signal. The motors in no case reach their peak velocities.

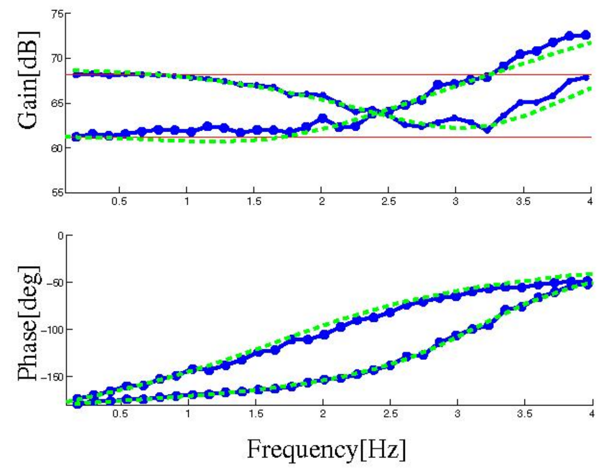

Fig. 6. Measured mechanical impedance of the system for the centre position (fig 4) for 1kN/m and 5 kN/m impedances settings. The dashed line shows the model based transfer The solid line represents the measurement. The measured static impedance gain was used as the spring stiffness of the model and the mass was derived using (4)

With a nearly straight leg it was possible to simulate impedances of up to 20 kN/m when perturbing the system manually. However this measurement could not be verified using the perturbation motor due to a lack of power as at this point the apparent mass of the system from the point of the actuator is very large. Impedances higher than 20kN/m were not possible due to the signal to noise ratio of the sensors. At these nearly straight orientation of the leg the system is able to support the entire bodyweight with the given moments. However when the system was in a position similar to the central position of fig 4 the moment of the system was not enough to offer enough virtual force between the ankle and the hip.

In the horizontal impedances the maximum mechanical impedance measured with a reasonable amplitude was 0.5kN/m. This limit was also caused by the maximum moment produced by the motors. Also at higher impedance settings the noise became an issue. This is due to the limited resolution of the joint angle sensors. (potentiometers).

Based on a third order approximation of the transfer function we have identified the mass, damping and impedance of the system. The estimated mass was compared to the expected

TABLE II

MODELLED MASS VS MEASURED APPARENT MASS AT ALL MEASURED GRIDPOINTS. (ALL MASSES IN KG)

	x=1		x=2		x=2	
	model	measure	model	measure	model	measure
$y = 1$	8	8.5	7	6.5	8	8
$y = 2$	10	10.5	11.5	12	10	9.5
$y = 2$	12.5	12	11	10	12.5	13

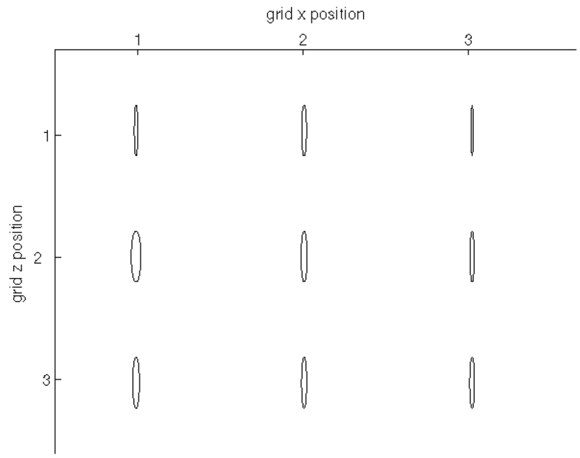

Fig. 7. This figure shows the ability of the system to offer a pure force/ impedance in the vertical direction. Each elipse represents an average over all trials measured at the grid position corresponding to the grid in figure 4. The height of each elipse shows the RMS value of the forces in the vertical direction and the width of the elipse shows the RMS value in the x direction.

mass (4) and the results are shown in table II.

Force Selectivity The exertion of forces in one purely one direction is an important measure for the ability to selectively support different gait functions while not influencing other functions. In fig 7 we can see the force selectivity per position when the system was performing a vertical impedance. A single elipse shows the average relationship for all measurements at that grid position between the RMS value of the vertical direction and the RMS value of the horizontal direction. In order to be able to take this average we normalized using the RMS for the vertical position. In an ideal case the elipses would be straight vertical lines, but in this case we can see that a small component is present in the horizontal direction.

IV. DISCUSSION

Zero impedance: Using the controller it was possible to decrease the impedance of the system by a factor of 10. The leg when loosed will swing out over several passes of the neutral angle. The measured impedance was dominated by the inertia of the system. This means that for slow movements like CVA gait in combination with the low inertia of the exoskeleton it is possible to effectively make the system transparent to the subject. Even at higher gait speeds the test persons did not feel hampered by the robot.

Vertical impedance: We have determined a maximum stiffness of 5 kN/m that can be guaranteed over the entire movement range. By choosing heavier motors or a higher transmission ratio it is possible to increase these impedances. The sensor noise was not a problem because at the exoskeleton orientations found during normal gait the vertical position measurement is not sensitive to signal noise is low. The applied DC motors could each deliver 25 Nm this should be at least 50 Nm in a next prototype. We have measured higher impedances at orientations of the leg bordering on the singular orientation (a stretch leg). However when the leg is flexed more than approximately 20 degrees at the knee the robot was unable to sustain adequate moments to guarantee the stiffness above 5 kN/m.

Horizontal impedance: The maximum impedance determined here was 0.5 kN/m at a frequency of 2 Hz as an ideal spring. At frequencies up to 4 Hz the system still performed predictably as a spring damper mass system. At higher stiffness settings we have found that the controller was too noisy at orientations close to the singular position. The arm between the hip and the ankle is maximum at the singular orientation making the controller more sensitive to measurement noise in of the hip angle measurement. This problem can be overcome by minimizing play in the joint and using more accurate sensors. In the next prototype the forward direction of the pelvis is also actuated. Using sensors placed in this axes reduces the effect of noise on the control system.

Force Selectivity: It was possible to exert forces in any direction without any major cross talk into other directions. The main bleed into different direction was caused by sensor noise at low frequencies and at higher frequencies the inertia. However the frequencies at which the mass plays a dominant role is far above the gait shown by stroke patients.

The results of this pilot measurements indicate that the system is capable of producing the necessary impedances in order to support gait given that the system is improved slightly over the prototype. Now trial studies are needed to show that it is possible to selectively support gait.

V. CONCLUSION

In this article we have shown that it is possible to implement a VMC using series elastic actuators with a non-linear transmission. For the vertical we measured an stiffness of 5kN/m for all ranges at frequencies of up to 1Hz as a near ideal spring. In the horizontal we measured op to 0.5kN/m in the same frequency range. The system shows predictable behavior for the entire movement range and frequencies present in stroke impaired gait for the given impedances. Also using a VMC type control it was possible to generate end-point forces selectively in one specific direction without a large degree of crosstalk into the orthogonal direction.

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