ELSEVIER

Gait and Posture 12 (2000) 94–104

GAIT & POSTURE

www.elsevier.com/locate/gaitpost

Postural activity of constrained subject in response to disturbance in sagittal plane

Matjaž Mihelj [a,*], Z. Matjačić [b], T. Bajd [a]

[a] *Faculty of Electrical Engineering, University of Ljubljana, Tržaška 25, 1000 Ljubljana, Slovenia*
[b] *Center for Sensory-Motor Interaction, Aalborg University Fredrik Bajers Vej 7 C-1, DK-9220 Aalborg, Denmark*

Accepted 3 May 2000

Abstract

This study examines the postural activity of partially constrained subjects for three different initial standing postures in response to disturbances in the sagittal plane. When the ankle strategy suffices for disturbance rejection in response to anterior disturbances, a mostly linear relationship between the ankle torque and ankle angle was observed, resulting in a constant stiffness at the ankles. However, when the ankle torque saturation was reached, a combined ankle-trunk postural strategy was utilized mainly in response to posterior perturbations due to the properties of the base-of-support. This caused a nonlinear scaling of ankle responses, thereby increasing variability of ankle stiffness. Distinctions in the ankle responses were also observed for different initial standing postures. The anterior initial stance generally increases the overall postural stability and renders the utilization of ankle strategy feasible, even for the rejection of posterior disturbances. Therefore, a linear torque–angle relationship at the ankles was observed for the anterior initial stance, regardless of the perturbation parameters. © 2000 Elsevier Science B.V. All rights reserved.

Keywords: Posture; Balance; Electromyograms (EMG); Ankle stiffness

1. Introduction

Arm-supported standing and limited crutch or walker assisted walking can be restored in patients with spinal cord injuries through functional electrical stimulation or mechanical bracing of paralyzed lower extremities [1]. For these patients, standing has many beneficial effects — both physiological and psychological in nature. Physically, standing may prevent joint contractures by interrupting the chronic sitting posture and may diminish osteoporosis. The upright posture may also improve functioning of internal organs and aid in bowel and bladder function. In addition, standing may provide significant functional assistance in accomplishing everyday activities. For example, a paraplegic patient would be able to reach some objects while standing that could not be reached from the confines of a wheelchair. These increased functional abilities may enhance personal self-esteem while providing a level of independence. Considerable effort is therefore being invested into finding an efficient methodology that would enable a paraplegic person to stand without arm support.

Unsupported standing has been achieved by bracing the paraplegic person's body above the shanks and applying electrical stimulation to the ankle joint plantarflexors [2,3]. The performance of the proposed control scheme is primarily limited by muscle fatigue and spasticity, which would cause falls under normal conditions. Even so, the controller might still be useful for freeing the hands during non-spastic and unperturbed periods. However, the efficiency of the proposed control scheme is mostly limited by the fact that only the artificial controller provides control signals for balancing, while there is no activity in the intact part of the paraplegic person's body.

A control strategy for unsupported paraplegic standing, utilizing the residual sensory and motor abilities of a thoracic spinal cord injured subject, was proposed by Matjačić and Bajd [4,5]. The strategy is based on

* Corresponding author. Tel.: + 386-147-68339; fax: + 386-147-68239.
E-mail address: matjaz.mihelj@robo.fe.uni-lj.si (M. Mihelj).

0966-6362/00/$ - see front matter
PII: S0966-6362(00)00065-5

voluntary and reflex activity of the patient's upper body and artificially controlled stiffness in the ankles. The knees and hips are maintained in the extended position by long leg braces or functional electrical stimulation. Thus, the subject is constrained in a double link inverted pendulum structure.

When assisted by an artificial ankle joint stiffness value of 8 Nm/deg or more, healthy and paraplegic subjects were capable of the proposed under-actuated balancing (under-actuated stands for a system that has less actuators than degrees of freedom, in this case the only actuator are the upper trunk muscles since there is no actuation in the ankle joints except the passive stiffness). Both healthy and paraplegic subjects were also capable of controlling balance when standing was perturbed with different anterior–posterior disturbances, indicating that constant artificial ankle joint stiffness enables efficient standing. Understanding the healthy subject's control over the constrained balancing might provide useful information that could improve the quality of the proposed control strategy further.

When a healthy subject is standing erect and the stance is imperceptibly perturbed, a linear relationship between the ankle torque and the ankle angle can be expressed as the load stiffness [6]. These small perturbations cause only negligible ankle angle fluctuations where ankle strategy alone suffices for maintaining the balance. Therefore, a simple single link inverted pendulum mechanical model of stance predicts the torque–angle relationship at the ankles. This relationship determines the minimal ankle joint stiffness required for a subject to stand unsupported. In the case of major disturbances, the ankle strategy by itself is not sufficient to reject the perturbation. The combined ankle–hip strategy is then required [7]. A similar result was observed in [5] where artificially controlled stiffness in the ankles did not satisfy the demands for overall postural stability. Leaning of the subject about the ankles prior to the disturbance also affects the overall postural strategy for disturbance rejection [8]. Subjects use a different postural strategy when perturbed while leaning near their forward or backward limits of stability. In this way, postural coordination changes in order to minimize muscle activation during different biomechanical conditions [9].

An important feature of postural dynamics is the effect of the forward lean. This lean results in a significant increase in the tonic component of ankle torque due to the increased muscle impedance. Nevertheless, the forward lean does not result in stiffening of the posture [10], but rather increases the postural stability by simplifying the response to the perturbation. Here the center-of-mass is located closer to the limits of the base-of-support facilitating the hip strategy [8]. At the same time, the risk of falling backward is reduced by increasing the stability margin between the center-of-gravity and the posterior limits of the base-of-support. Thus, the postural ankle dynamics are based on a single muscle group-ankle plantarflexors. This may allow simplification of artificial ankle muscle control in paraplegic patients. On the other hand, a backward lean results in decreased postural stability. It is therefore important to assess the responses to disturbances in similar situations.

According to the analysis of the perturbed stance explained above, the following hold true when subjects are standing unconstrained:

- when the stance is imperceptibly perturbed a linear relationship between the ankle torque and ankle angle was observed, thereby resulting in constant ankle stiffness;
- when the intensity of the perturbations increases, the stance posture is changed to an anterior lean in order to increase the stability margin and simplify the response.

It was our aim to investigate these points of interest in the performance of healthy subjects that were standing braced in a manner identical to paraplegic subjects in [5]. We were particularly interested in the performance of the ankle joints after the action of perturbations in order to find an appropriate control for the paralyzed muscles. The study addresses:

- the assessment of responses to different intensities of disturbances in the sagittal plane when a healthy subject is constrained in a double link inverted pendulum structure;
- the assessment of relationship between the ankle joint angle and the corresponding ankle joint torque, and
- the investigation in how human postural responses are altered by leaning about the ankles to three different initial stance positions prior to the onset of the disturbances.

2. Methods

Eight healthy male subjects aged 23 ± 4 years, with a weight range of 75 ± 10 kg and a height range of 176 ± 5 cm voluntarily participated in the investigation.

2.1. Experimental set-up

The experimental set-up is based on a mechanical rotating frame device [5]. Fig. 1 shows the (a) diagonal and (b) lateral view of a subject standing constrained in the mechanical rotating frame (MRF). The device consists of a base and a rotating frame. The base consists of a steel plate, bearings, and a hydraulic actuator. The rotating part of the device is a frame that provides bracing to the subject's lower body. The knees and hips are forced into an extended position by aluminum bars.

Two vertical, one posterior horizontal, and two anterior horizontal bars prevent the subject's knees and hips from moving. All horizontal bars are covered with a soft material. The fourth bar is placed behind the heels of the subject in order to prevent the feet from moving backward.

The rotating section of the device weighs about 15 kg, and its center-of-gravity lies about 0.2 m above the rotation axis. The total inertia of the rotating frame about the rotation axis is approximately 3 kg m^2. The influence of the mechanical rotating frame on the subject's balancing capabilities should not be significant since the rotating frame does not add much to the inertia of the subject's lower body (about 18 kg m^2 [5]).

The hydraulic subsystem provides the torque required to produce the various disturbances. The subsystem consists of a hydraulic pump, a servo valve, a hydraulic actuator, and two pressure transducers. The disturbance is then implemented as a pulse shaped torque (40 ms rise time from 5 to 95% of torque amplitude) generated by the hydraulic actuator that causes the frame to rotate about the subject's ankle joint axis. The perturbation torque is transformed via the rotating frame into a force acting on the subject's pelvis through the top two horizontal bars. The details on MRF are given in [5].

2.2. Visual feedback implementation

In order to enable the subject to assume the required initial stance position prior to the disturbance commencement, a visual feedback was provided to the subject. An animated double link inverted pendulum, representing the constrained subject's current posture, was presented on a screen one meter in front of the subject. Alongside the pendulum, the boundaries of the initial stance posture were also displayed. The subject was required to assume a posture such that the animated inverted pendulum remained between the boundaries. These limits were set to $\pm 0.3°$ and $\pm 0.6°$ off the selected posture for the lower and upper body, respectively.

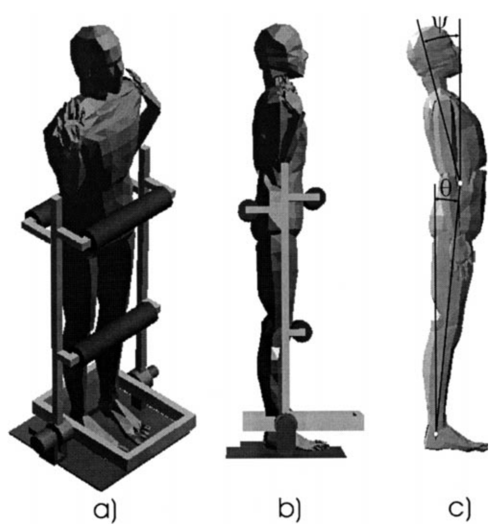

Fig. 1. Subject standing in MRF: (a) diagonal view; (b) lateral view and (c) angle definitions.

2.3. Experimental conditions

After the posterior horizontal bar was removed, the subject entered the mechanical rotating frame from the rear. The pelvis and feet were positioned in such a way that the lumbosacral joint axis (the spinal column was simplified as a single joint located at vertebrae L5-S1) and the ankle joint axis intersected the midline of the vertical bar of the frame. The ankle joint axis was also aligned with the axis of the frame bearings. The subject was constrained to allow movement only in his upper trunk and ankle joints, while the arms were folded on his chest.

The kinematics were assessed by the optical position measuring system OPTOTRAK® (Northern Digital Inc.). Two infrared markers were attached to the rotating frame, one on the bearings axis and the other on the vertical bar of the bracing system at the height of the subject's lumbosacral joint axis. Two additional markers were attached to the subject's trunk with the first located on the midline of the rib cage half way between the iliac crest and the shoulder. The second marker was then located five centimeters below the first. The assessment of the upper trunk inclination was rather poor due to the simplified representation of the spinal column as single joint and due to unreliable upper trunk marker position.

The joint angles were defined (see Fig. 1c) as follows:

- the ankle joint angle θ represents the angle between the lower extremities and the vertical axis (positive angles correspond to ankle dorsiflexion);
- the trunk inclination Ψ represents the angle between the upper trunk and the vertical axis (positive angles correspond to trunk flexion).

The base of the mechanical rotating frame was firmly attached to the force plate (AMTI, Advanced Mechanical Technology Inc.), measuring the reaction forces and torques during the experiments.

The electromyograms (EMG) of the triceps surae muscles and the tibialis anterior muscles of the right leg were recorded in order to provide an insight into the disturbance rejection strategy from the ankle muscle activation point of view and to assess the latency between the onset of the perturbation and the subject's response. Precision differential amplifiers (frequency band 50–50 000 Hz, gain 5000) were used to preprocess the EMG signals. The sampling rate for all signals was 500 Hz. The EMG signals were full-wave rectified and

Table 1
Perturbation parameter (I is the subject's inertia around the ankles)

Pert. type	Torque (Nm)	Duration (ms)	Average power (W)	Average energy (J)
1: Anterior & posterior	30	150	$67.5/I$	$10.0/I$
2: Anterior & posterior	30	250	$112.5/I$	$28.1/I$
3: Anterior & posterior	50	150	$187.5/I$	$28.1/I$
4: Anterior & posterior	50	250	$312.5/I$	$78.1/I$

lowpass filtered in both directions, thus preserving the phase content of the signals.

2.4. Experimental protocol

Three postures were tested: (1) a normal lean angle where the subject felt most comfortable; (2) approximately 3° anterior to the normal lean angle, and (3) approximately 3° posterior to the normal lean angle. Responses to eight different types of perturbation were measured. The perturbations differed in direction, torque pulse amplitude M_{pert}, and pulse duration time T_{pert}. The parameters are summarized in Table 1.

The experimental protocol was based on a pilot study with three subjects, which involved results that were presented elsewhere [11]. The smallest perturbation was determined to cause a barely noticeable response, while the largest perturbation was selected in such a way that it did not cause the subject to fall or make a step. The average disturbance power P_{pert} and energy W_{pert}, the values of which are computed from a single link inverted pendulum model, are presented in Table 1. The perturbation energy was estimated from the following equation

$$W_{pert} = \frac{M_{pert}^2}{2I} T_{pert}^2 \qquad (1)$$

where I stands for the subject inertia around the ankle joints. The average power was then computed as a ratio between the perturbation energy and the pulse duration time

$$P_{pert} = \frac{M_{pert}^2}{2I} T_{pert} \qquad (2)$$

However, because the subject's body breaks in a double link inverted pendulum structure after the onset of the perturbation that results in a lower inertia, the estimated values from Eq. (1) and Eq. (2) indicate the minimum of the disturbance average power and energy. Nevertheless, the simplified computation of the average power and energy gives some approximate values that enable comparison between different perturbation types.

The experiments began with a series of eighty familiarization trials. The disturbances were selected in random order, and the subject was encouraged to assume different postures prior to the disturbance implementation. Prior to the testing phase, the subject was instructed to assume his most comfortable standing posture and relax. This posture was recorded as the normal lean angle. Afterwards the subject was asked to keep his arms folded on his chest and focus his attention on the visual feedback that provided information on the current posture and the required initial stance margins. In order to assure that the subject was relaxed prior to the perturbation, the EMG signals were displayed on an oscilloscope. The disturbance was not initiated until the muscular activity decreased below the value initially assessed as the resting muscular activity.

The testing phase was divided into three subphases. Each subphase was associated with a different initial stance posture in the following order: normal lean, anterior lean, and posterior lean. The subphases consisted of eighty trials grouped into ten groups of eight. Each group was a sequence of all perturbation types implemented in random order. Average time courses of segment angles and joint torques were computed based on the results from ten trials for each experimental condition. The correlation coefficient of the linear regression for ankle torque versus ankle angle relationship was determined for each trial. The average ankle stiffness for each trial was then determined as the slope coefficient of the ankle torque versus ankle angle relationship by applying the linear regression technique.

3. Results

3.1. Kinematics

Fig. 2 shows the average time courses of the ankle joint angle based on ten trials of subject No. 8 in response to various disturbances executed at different initial stance postures. The onset of the disturbance is set to time 0 s. The initial stance ankle joint angle values were approximately 2° for normal lean, 5° for anterior lean, and −1° for posterior lean. The plots indicate low variability of the initial stance postures, thereby enabling adequate comparison of results. After the onset of the perturbation, the ankle joint angle was displaced in the direction of the perturbation. It can be observed that the amplitude of the ankle joint angle displacement varies with the type of the induced perturbation and the initial stance posture. However, the results indicate that the initial stance posture primarily affects the strategy of the response to posterior disturbances. The amplitude of the ankle joint angle displacement for the anterior disturbances is mainly related to the type of the disturbance and almost independent of the pre-perturbation posture, while the initial stance posture considerably affects the postural dynamics following the posterior disturbances. The principal reason for the observed differences in these responses can be found in different postural strategies that have to be adopted due to the biomechanical constraints. While the subject was able to reject all anterior disturbances relying mainly on the ankle strategy with only small activity of the upper trunk (Fig. 3), the situation changes considerably with posterior perturbations. The trunk response increases by increasing the magnitude of the perturbation and by moving the initial stance posture backward. Therefore, the largest trunk flexion can be observed in the case of the largest posterior disturbances at the posterior initial stance. The short pulse shaped response of the upper trunk inclination at anterior lean was mainly a consequence of the trunk's inertia and was promptly compensated, adding only small contribution to the overall postural strategy.

As mentioned previously, the ankle joint response is mainly related to the magnitude of the perturbation. The smallest perturbation resulted in a minimal displacement, and the largest perturbation resulted in a maximal displacement. Both results were as expected. On the other hand, the responses to the perturbations of types two and three are almost identical. The second perturbation type has a longer duration (250 ms) with lower amplitude (30 Nm), and thus a lower average power, while the third perturbation type is short (150 ms) with higher amplitude (50 Nm), and thus a higher average power. It would therefore seem that the perturbation power is not a relevant parameter. However, the energy induced into the system by both perturbations is almost identical, indicating that the total energy better describes the characteristics of the disturbance in view of the responses.

Two distinct kinematic responses can be observed in the sagittal plane, depending on the perturbation direction. The anterior disturbance caused a simultaneous increase of the ankle joint angle and a decrease of the upper trunk angle due to the trunk inertia. A fast trunk response returned the upper body to an erect posture in approximately 500 ms. During this time, a prominent ankle action resulting in deceleration of the overall body movement was observed. Finally, the continued ankle response returned the whole body to the vertical posture. At the same time, small corrections of the upper trunk can also be observed.

Posterior perturbations, however, caused a simultaneous decrease of the ankle joint angle and increase of

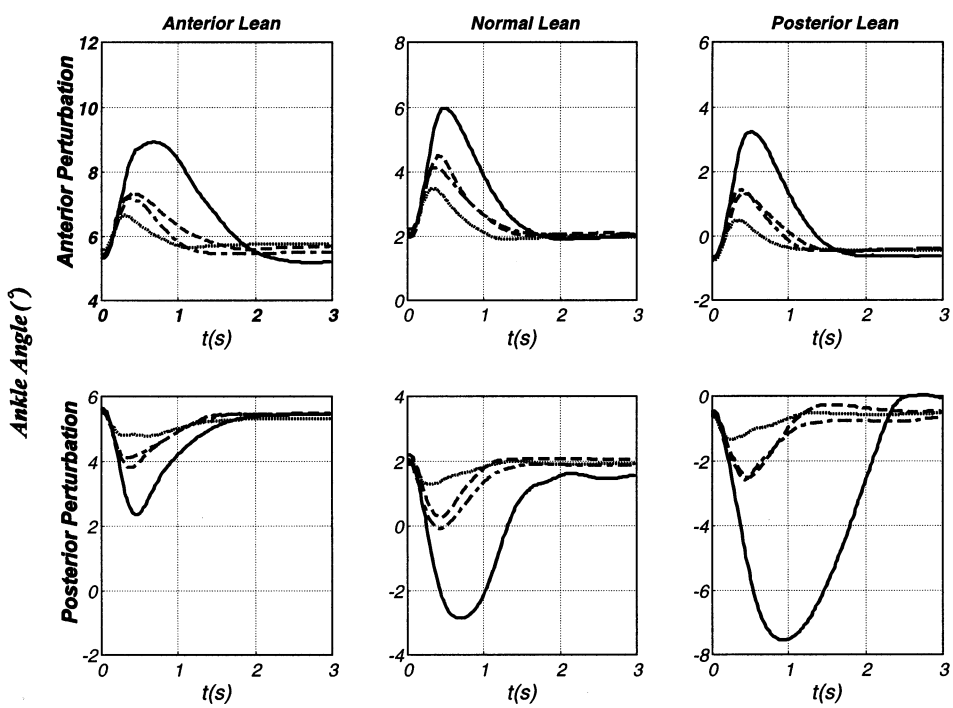

Fig. 2. Ankle angle for different experimental conditions based on averaging over ten trials for each experimental condition (subject No. 8). The onset of the disturbance is set to time 0 s. Different line styles indicate different perturbation types according to Table 1: type 1 ⇒ dotted line, type 2 ⇒ dash-dotted line, type 3 ⇒ dashed line, type 4 ⇒ solid line.

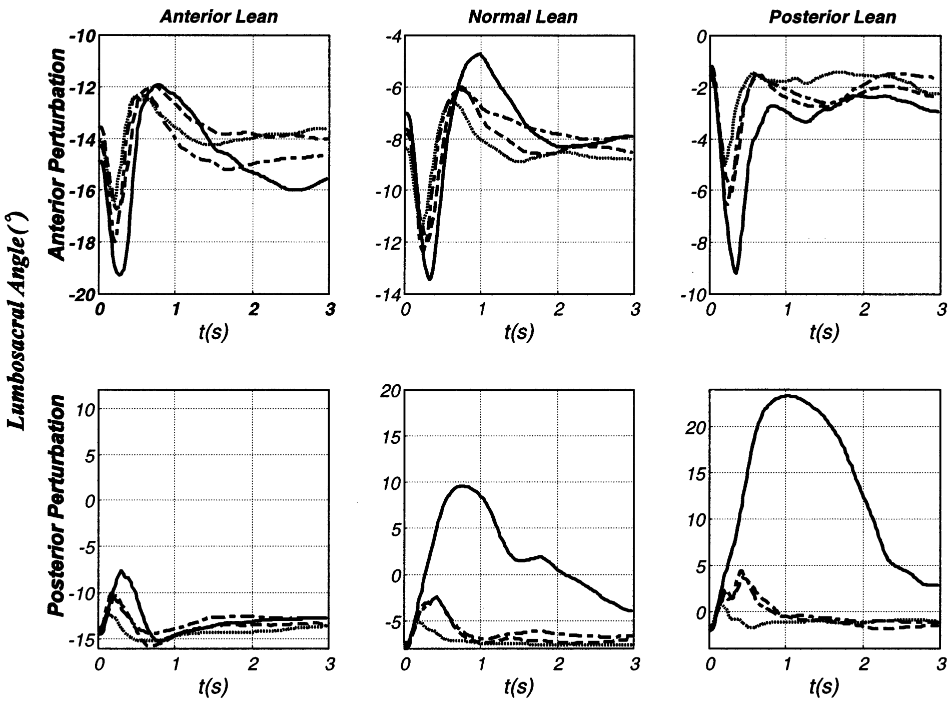

Fig. 3. Upper trunk inclination for different experimental conditions based on averaging over ten trials for each experimental condition (subject No. 8). The onset of the disturbance is set to time 0 s. Different line styles indicate different perturbation types according to Table 1: type 1 ⇒ dotted line, type 2 ⇒ dash-dotted line, type 3 ⇒ dashed line, type 4 ⇒ solid line.

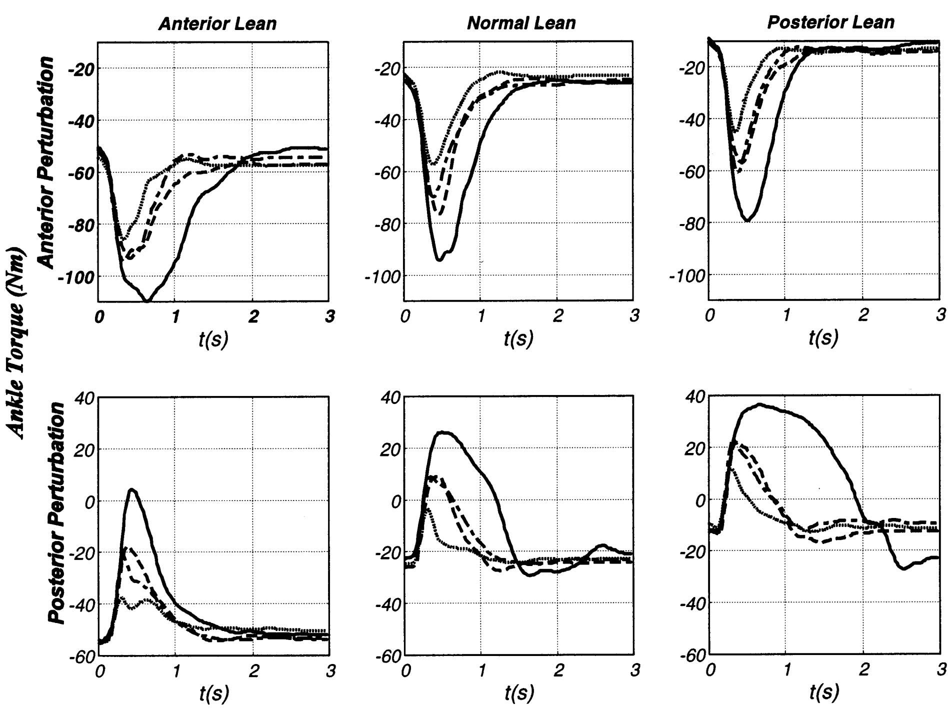

Fig. 4. Ankle torque for different experimental conditions based on averaging over ten trials for each experimental condition (subject No. 8). The onset of the disturbance is set to time 0 s. Different line styles indicate different perturbation types according to Table 1: type 1 ⇒ dotted line, type 2 ⇒ dash-dotted line, type 3 ⇒ dashed line, type 4 ⇒ solid line.

the upper trunk angle due to the trunk inertia. Since the ankle joint strategy was not sufficient to compensate for the disturbance, the trunk flexion continued to provide fast anterior shifting of the center-of-mass in order to stabilize the body movement [12]. In the next stage, a simultaneous inversion in both joint movements occurred, finally returning the whole body to a vertical stance.

3.2. Dynamics

Fig. 4 shows the ankle torque as a response to the perturbation. The ankle torque is the sum of the contributions of both ankles. The ankle torque during the quiet stance (time 0 s), the tonic component, is related primarily to the initial stance posture. As a consequence of the perturbation, the ankle torque is changed in order to oppose the change in the ankle joint angle. The amplitude of this ankle joint torque change varies with the type of induced perturbation and the initial stance posture. It can be noticed that the amplitude of the torque change in the ankle joint increases by increasing the total perturbation energy, which is consistent with the changes in the ankle joint angle. However, on average, the results indicate smaller amplitudes of the torque change in the ankle joints for the posterior disturbances as compared to anterior disturbances.

3.3. EMG responses

Fig. 5 shows the characteristic time signatures of the EMG signals for the tibialis anterior and triceps surae muscles. EMG signals were normalized by assigning a value of one to the activated muscle in the normal stance position for the anterior sway (triceps surae) and for posterior sway (tibialis anterior). Values for other initial stance positions were then expressed as a ratio [8].

The tonic muscular activation depends solely on the initial stance posture. Prominent tonic activation of the triceps surae muscles can be observed in the case of an anterior lean. This tonic activation provides enough muscular torque to compensate for the effect of the increased gravity. The phasic muscular activity is highly related to the type of the perturbation. When an anterior disturbance was applied, the triceps surae muscles became highly activated while almost no activity in the tibialis anterior muscles was observed. The exception occurred during a forward lean, where a small co-activation of these muscles was observed. In the case of posterior disturbances, the tibialis anterior muscles performed the majority of the disturbance rejection tasks. When the subject assumed the anterior initial stance, posterior perturbation caused a decrease in the tonic activity of the triceps surae muscles and a small phasic activation of the tibialis anterior muscles. In the other two initial stances, only a small activation of the triceps surae muscles was observed.

The onset latencies of the EMG bursts were measured from the first burst increase over the average background activity in the individual trials. The latencies were defined as the earliest time a sustained burst of EMG activity deviated from the preperturbation level of EMG activity and expressed with respect to the disturbance onset. [8]. The average tibialis anterior muscle

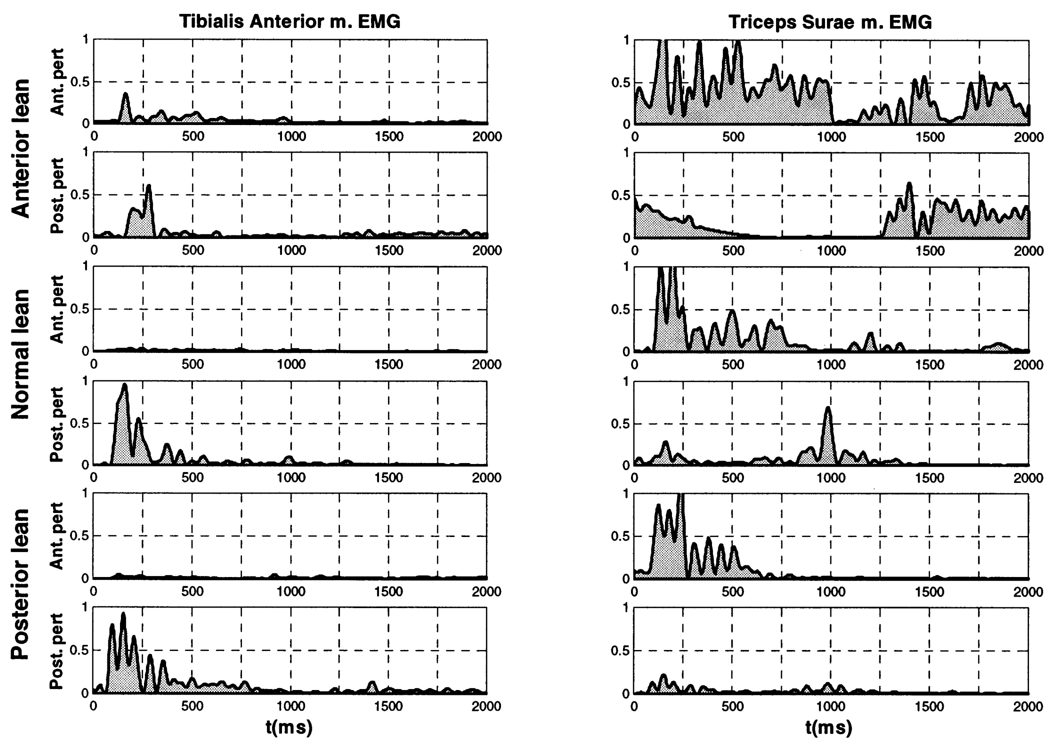

Fig. 5. Normalized characteristic time signatures of the EMG signals of tibialis anterior and triceps surae muscles for different initial stance postures and perturbation directions (subject No. 8, perturbation type 3). The onset of the disturbance is set to time 0 s.

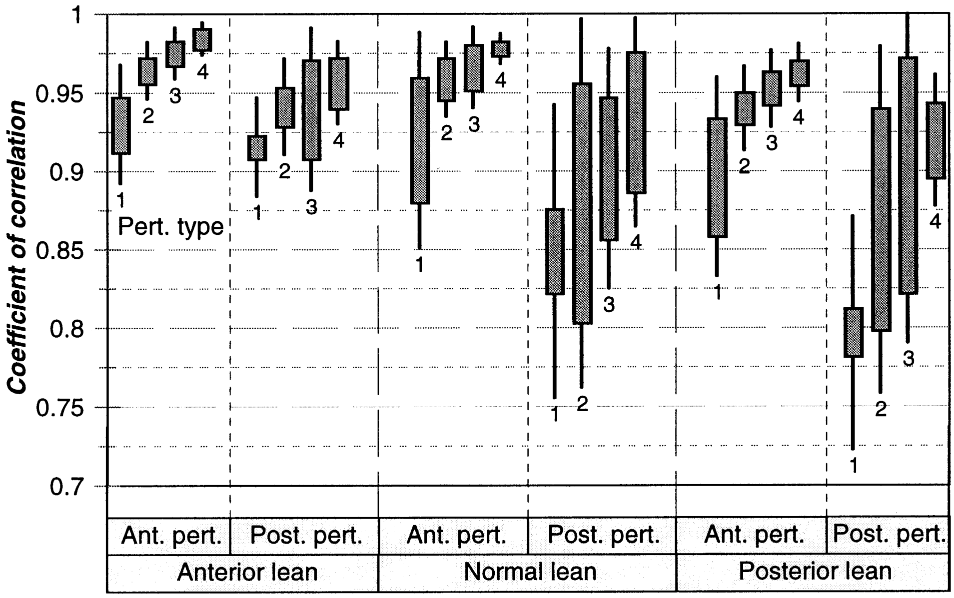

Fig. 6. The coefficient of correlation for all experimental conditions for 7 subjects. The results for the subject with the largest deviations from the average were eliminated. The shaded bars indicate the region, where the average coefficient of correlation values (based on ten trials for each subject) lie. The I bar on the bottom and top side of dark shaded bar indicates one standard deviation within ten trials for each subject averaged across all subjects.

latencies were found 105 ± 20 ms, while the average triceps surae latencies were 100 ± 15 ms for all perturbation types.

3.4. The relationship between ankle angle and ankle torque

Figs. 2 and 4 indicate a linear relationship between the ankle joint angle and the corresponding ankle joint torque.

In order to evaluate the goodness-of-fit for this relationship, the correlation coefficient of the linear regression values for seven subjects are plotted in Fig. 6. Results from the subject with the largest deviations from the average, which resulted from noticeably different initial stance posture, were eliminated. The shaded bars indicate the region where the average coefficient of correlation values lie. These average values are based on ten trials for each subject. The I bar on the bottom and top sides of the dark shaded bar indicates the

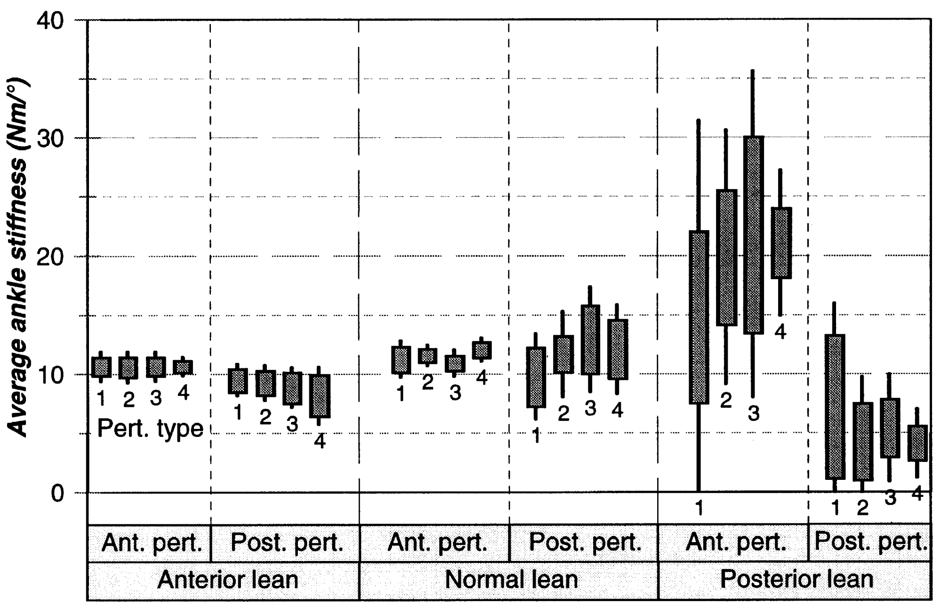

Fig. 7. The average ankle stiffness for all experimental conditions for 7 subjects. The results for subject with the largest deviations from the average were eliminated. The shaded bars indicate the region, where the average stiffness values (based on ten trials for each subject) lie. The I bar on the bottom and top side of dark shaded bar indicates one standard deviation within ten trials for each subject averaged across all subjects.

standard deviation for each subject averaged across all subjects for the ten trials. A predominantly linear relationship can be observed for all anterior disturbances, when the ankle strategy suffices for perturbation rejection. In contrast, when the posterior disturbances were applied in normal and posterior lean, the relationship between the ankle joint angle and the ankle joint torque deviated slightly from the linear one. These deviations increase when the subject is forced to rely more on the combined ankle-trunk or trunk-alone strategy.

Since the results indicate a prevailing linear relationship between the ankle joint angle and the corresponding ankle joint torque during the compensation of the disturbances, a notation of the ankle joint average stiffness was introduced. The ankle joint average stiffness represents the negative of the ratio between the ankle joint torque, produced in the subject's ankle joints in order to reject the disturbance, and the corresponding ankle joint angle. The results presented in Fig. 7 show the average ankle stiffness for seven subjects over ten trials. Again, results for the subject with the largest deviations from the average were eliminated, and the shaded bars indicate the region where the average stiffness values lie. The I bar on the bottom and top sides of dark shaded bar indicates one standard deviation based on ten trials for each subject averaged across all seven subjects.

The average ankle joint stiffness value for anterior and normal initial stance postures was found to lie between 9 and 12 Nm/deg, regardless of the perturbation type or subject. A detailed analysis of the average stiffness shows that differences between subjects were almost negligible for anterior and normal initial stance postures when anterior perturbations were applied. The stiffness value was found to lie between 10 and 11 Nm/deg. Relatively constant average stiffness values can also be observed for anterior initial stance and posterior disturbances, although the average is slightly lower (about 9 Nm/deg) than for the anterior disturbance cases.

Analysis of the average stiffness for a normal stance with posterior perturbations shows an increased variability between the subjects with the average stiffness 12 Nm/deg. This variability increase is most probably due to a changeover in the postural strategy from the ankle to the combined ankle-trunk strategy when the perturbation intensity is increased. During the transitions between these postural strategies, the inter-subject variability increases due to the differences in dexterity and anthropometric characteristics of the individual subjects.

The relations are markedly different when subjects assume a posterior initial stance posture. The ankle stiffness changes considerably, and the variability between the subjects' responses increases. The anterior disturbances resulted in an increased average stiffness value, while the average stiffness substantially decreased when the posterior disturbances were applied.

4. Discussion

The analysis of the postural control system in this paper was based on the perturbed standing of partially braced subjects. In order to study the postural control system, a device was used that provided bracing to the lower extremities and enabled implementation of different disturbances in the anterior–posterior direction. The motion constraints were designed to resemble the stance of a paraplegic patient with an artificially controlled degree of freedom in the ankles and the second degree of freedom in the upper trunk under voluntary and reflex control. The main difference between using moving platforms and our approach of generating disturbed standing is that the disturbance in our case is induced by pushing the subject in the height of the pelvis, which causes the biomechanical structure to break into two segments that are accelerated in opposite directions. The lower braced part of the body is accelerated in direction of the perturbation, while the upper trunk, due to its inertia, is accelerated in the opposite direction of the perturbation.

Such constrained standing enables the continuum of postural strategies based on the combination of ankle and trunk activity in order to obtain postural equilibrium. In comparison to unconstrained standing, the upper body activity is constrained to the trunk action since the hip joints are in full extension. Therefore, the hip strategy as defined in [9] becomes a trunk strategy in this case, which is less flexible with a predominantly anterior sway due to the biomechanical properties of the spinal column. The upper trunk contribution to the overall postural strategy is thereby effectively disabled. The transition from the ankle to trunk strategy occurs when the horizontal perturbations of the body's center-of-mass exceed a particular distance or velocity boundary for effective use of the ankle strategy. This is also the case with unconstrained standing or standing on a narrow beam [7].

The responses to applied disturbances are complex dynamic processes, occurring in a multi-segment mechanical system where different forces and torques act on these segments and the joints. In order to understand the process of balancing, it is necessary to have a model that describes the mechanics of the responses and also takes the complex adaptive postural control system into account. However, by eliminating less important parts of a model, the model becomes simpler and more analyzable. Yet at the same time also more abstract and distinct from the real system. Simplification of the model by using the linear relationship between the ankle torque and angle may be justified by considering the overall goal of finding a suitable control of paralyzed ankles to assist the paraplegic patients exercising arm free standing. Recall that this linear relationship indicates constant joint stiffness.

The energy resulting from the perturbation of a subject is, according to the amplitudes of postural responses, the most relevant parameter of the disturbance. The amplitudes of the responses generally increase with the energy induced by the disturbance. However, while the responses to anterior disturbances are mostly proportional to the energy level, the posterior perturbations result in nonlinear scaling of the responses. These findings may reflect nonlinear scaling of acceleration-dependent changes in postural strategy, resulting in increased use of trunk rather than ankle muscles [13]. This would also explain the nonlinear scaling of the responses that primarily occurs for posterior disturbances since in this case an extensive trunk flexion is possible due to the biomechanical properties of the spinal column. Nevertheless, the increasing involvement of the trunk muscles begins only when the perturbation magnitude reaches a certain threshold. This brings the ankle torque to its upper boundaries as determined by the length of the base-of-support posterior to the ankle joint axis. On the other hand, the anterior disturbances cannot be compensated for by extensive trunk activation due to the mechanical constraints of the spinal column and the extension of the hip joints provided by the bracing system. However, the rejection of the anterior disturbance is possible due to the biomechanical properties of the foot, which provides enough support for the ankle muscles to generate plantar flexion torque and thus stabilize body movement induced by the disturbance.

The results presented in this paper indicate that the posture leaning in the anterior direction provides the most adequate conditions for the disturbance rejection. The forward lean assures that the center-of-mass remains anterior to the ankle axis during posterior disturbances, thereby facilitating the response and diminishing the requirements for the trunk activation. Such leaning was observed in healthy subjects during continuous perturbations, particularly at higher perturbation magnitudes, and also when these subjects stood blindfolded [13]. Forward leaning reduces the need for dorsiflexor activation and, at the same time, increases the effective stiffness of the plantarflexors through changes in muscle mechanics and stretch reflex excitability due to increases in mean activation level [14]. In addition, subjects tend to lean forward in order to minimize the risk of falling backward. The resulting positive influence of the anterior lean on postural stability can also be observed in our experimental results where, even for powerful posterior disturbances, the threshold for trunk activation has not been exceeded. The response amplitudes in this case were generally proportional to the perturbation energy. The anterior lean therefore results in low variability of ankle responses that can be reliably described by constant stiffness.

As the initial stance posture is moved backward towards the normal stance, and subsequently further towards the posterior stance, the conditions for disturbance rejection deteriorate. The horizontal component of the COG approaches the ankle axis, and the body weight is insufficient to compensate the posterior disturbances. Therefore, the dorsiflexor action increases. As the posterior perturbation magnitude increases, the ankle torque reaches the saturation level due to the finite length of the base-of-support. (In our nomenclature the saturation level is not related to the maximal force the muscles can produce, but rather to the biomechanical constraints which result in toe-off or heel-off response.) Thus, the trunk flexion is required in order to move the center-of-mass anteriorly and stabilize the body. The ankle saturation level is reached sooner if the initial stance posture is moved backward. From the point of the saturation on, the nonlinear scaling of the responses becomes obvious due pronounced trunk contribution to the overall postural activity.

However, when the ankle torque is not saturated, the balancing is mostly based on ankle strategy. A single inverted pendulum model can now be used to describe the conditions. The literature describes several studies of postural control with a fully constrained body having only one degree of freedom in the ankles. Fitzpatrick et al. [6] reported a linear relationship between the ankle torque and the ankle angle for small perturbations and a voluntary sway in the sagittal plane. The average ankle stiffness value, when all sensory cues were available, was about 2 to 3 Nm/deg higher than our findings. However, this can be attributed to subject constraint [6] to one single postural strategy where the trunk movement was not possible.

The identification approach presented in this paper provides a new insight into the problem of balance control for subjects constrained in such a way that balancing strategy is based only on the ankle and trunk muscles. The responses to small anterior disturbances were based on the ankle strategy, while larger disturbances resulted in a combined ankle-trunk strategy.

Based on the computation of the coefficient of correlation, the results of the study indicate a predominantly linear relationship between the ankle joint angle and the corresponding ankle joint torque for a series of initial stance postures and perturbation intensities. The value of the linear coefficient representing the average ankle stiffness, defined as the ratio between the ankle torque and ankle angle, was found to be constant for the significant part of experimental conditions.

The results show that the most adequate ankle stiffness ranges between 9 and 12 Nm/deg. This is in a substantial agreement with the findings of a simulation-based study [4], indicating that the constant ankle stiffness value around 10 Nm/deg provides the most adequate conditions for the disturbance rejection. Such stiffness values provide good conditions for disturbance rejection in the sagittal plane when the initial stance posture lies between an anterior and a normal lean. On the other hand, in the case of a posterior lean, the variability of the results is increased. Therefore hardly any conclusions about the most appropriate stiffness can be drawn for this condition. However, since the strategy of the response to a posterior lean becomes more complex with the prevailing upper trunk activity, the contribution of the ankles to the overall postural strategy diminishes. This increases the set of adequate ankle stiffness values and can also explain the high variability of the results for the posterior initial posture.

It should be emphasized that although the average ankle stiffness was used to describe the postural control during perturbed stance, the choice of stiffness as the
during perturbed stance, the choice of stiffness as the
measured variable does not imply that the nervous system primarily regulates the joint stiffness. However, the presented results support the choice of simple stiffness control in the paralyzed ankles for a standing, unsupported paraplegic subject. This is based on voluntary and reflex activity of the paraplegic person's upper body and artificially controlled constant stiffness in the ankles as proposed in [4] and [5].

References

[1] Kralj A, Bajd T. Functional Electrical Stimulation: Standing and Walking After Spinal Cord Injury. Boca Raton, FL: CRC Press, 1989.

[2] Hunt KJ, Munih M, Donaldson N de N. Feedback control of unsupported standing in paraplegia: part I: optimal control approach. IEEE Trans Rehab Eng 1997;5:331–40.

[3] Munih M, Donaldson NdeN, Hunt KJ, Barr FMD. Feedback control of unsupported standing in paraplegia: part II: experimental results. IEEE Trans Rehab Eng 1997;5:341–52.

[4] Matja(i(Z, Bajd T. Arm-free paraplegic standing: part I: control model synthesis and simulation. IEEE Trans Rehab Eng 1998;6:125–38.

[5] Matjačić Z, Bajd T. Arm-free paraplegic standing: part II: experimental results. IEEE Trans Rehab Eng 1998;6:139–50.

[6] Fitzpatrick RC, Taylor JL, McCloskey DI. Ankle stiffness of standing humans in response to imperceptible perturbation: reflex and task-dependent components. J Physiol 1992;454:533–47.

[7] Horak FB, Nashner LM. Central programming of postural movements: adaptation to altered support-surface configurations. J Neurophysiol 1986;55:1369–81.

[8] Horak FB, Moore SP. The effect of prior leaning on human postural responses. Gait Posture 1993;1:203–10.

[9] Nashner LM, McCollum G. The organization of human postural movements: a formal basis and experimental synthesis. Behav Brain Sci 1985;8:135–72.

[10] Sinha T, Maki BE. Effect of forward lean on postural ankle dynamics. IEEE Trans Rehab Eng 1996;4:348–59.

[11] Mihelj M, Matjačić Z, Bajd T. Assessment of intact subject's balancing strategy. Proceedings of the 6th Vienna Workshop on Functional Electrostimulation, Vienna (Austria), September 22–24, 1998, 129–132

[12] Henry SM, Fung J. Control of stance during lateral and anterior–posterior surface translations. IEEE Trans Rehab Eng 1998;6:32–42.

[13] Maki BE, Ostrovski G. Scaling of postural responses to transient and continuous perturbations. Gait Posture 1993;1:93–104.

[14] Maki BE, Whitelaw RS. Influence of expectation and arousal on center-of-pressure responses to transient postural perturbations. J. Vestib. Res. 3 25–39