Arm-Free Paraplegic Standing—Part II: Experimental Results

Zlatko Matjačić and Tadej Bajd, *Senior Member, IEEE*

***Abstract*— In Part I, we proposed an approach for restoring unsupported standing to thoracic-level paraplegics. The theoretical analysis and simulation of an underactuated double inverted pendulum, representing the standing subject, showed that arm-free standing might be achieved. Here in Part II, we present the mechanical apparatus which we used in our experiments and experimental results from tests of the balance-control strategy. We demonstrate that an intact and a paraplegic subject could perform quiet standing with the ankle stiffness set to 8 Nm/° or even less (the intact subject). Both were also able to recover from disturbances, imposed by the artificial ankle joint of the apparatus. Introducing cognitive auditory feedback greatly improved the standing abilities of both subjects.**

***Index Terms*—Mechanical rotating frame, natural and artificial control of standing, paraplegia, voluntary and reflex balancing.**

I. Introduction

A. Problem Statement

In Part I, we proposed a control strategy for arm-free paraplegic standing. Through theoretical analysis and simulations, we have shown that with a properly selected artificial stiffness in the ankles, paraplegic subjects should be able to stand by reflex and voluntary activity of their preserved trunk muscles but without using their arms.

While it is our long-term aim to test this strategy in experiments, using FES to maintain the knees and hips in full extension and to control the ankle stiffness (Part I, Fig. 2), the use of FES may lead to complicated behavior due to fatigue of the stimulated muscles, spasms and ankle spasticity. There will also be day-to-day variability due to the dependence of muscle force on the electrode position. Eventually the effects of these complications on the behavior will have to be assessed. However, we can divide the experiments into two or more stages. In the first stage, we want to know whether a paraplegic can balance if his legs behave approximately like an ideal lower link of the double-pendulum. Not only should the knees and hips be held in extension and the ankle stiffness be accurately controlled, but also, as we are interested in sagittal plane stability, lateral motion should be prevented. Only if the experiments in the first stage are successful, would there be any point in progressing to later stages, in which FES will be used and lateral motion possible. This paper describes the experiments in the first stage.

Due to these considerations, an apparatus named the mechanical rotating frame (MRF) was designed and built. The MRF guarantees the desired behavior of the lower link of a double inverted pendulum, it braces the knee and hip joints in extended positions and both ankle joints are constrained in a neutral position. The single rotational degree of freedom of the apparatus functions as an artificial ankle joint. In this way, the MRF fulfills the requirements for repeatable behavior of the lower link, lateral stability and safety, as will be demonstrated in the following section. It also enables investigation of the proposed control strategy in intact subjects because the natural leg joints are braced.

There have been two studies that investigated a role of cognitive feedback in paraplegic standing. Turk *et al.* [2] demonstrated the use of cognitive feedback which may prolong the standing time in paraplegics using FES and single-arm support. Phillips and Petrofsky [3] showed that cognitive feedback information could enable a paraplegic to balance in an orthosis (RGO) without any arm support. Even though the theoretical study from Part I suggested that the vestibular and visual systems might be sufficient for successful balancing, we also showed that a slight change in the inclination of the lower link (2°) of the pendulum considerably reduced the volume of the feasible perturbation space in case of posterior disturbances (Part I, Fig. 11). Therefore, it is likely that a cognitive feedback would enhance performance. In addition to vestibular and visual information (during the present experimental investigation) auditory sensory feedback, communicating the inclination of the lower part of the body, was also provided to the standing subject.

B. Objectives

The purpose of the experimental investigation, presented in this paper, was exploratory. The main question we wanted to answer was whether intact and paraplegic subjects are capable of arm-free standing, utilizing the proposed control strategy with and without cognitive feedback. The second question was how both subjects behave when standing was perturbed by means of torque impulses imposed by the artificial ankle joint. We were particularly interested in the central neural system (CNS) delay and the strategy used to recover from disturbances. We also studied the influence of vision and artificial cognitive feedback by blindfold standing trials. In the Introduction to Part I where we proposed the control scheme, we stated that the CNS of a subject should be able to cope

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The authors are with the Faculty of Electrical Engineering, University of Ljubljana, 1000 Ljubljana, Slovenia.

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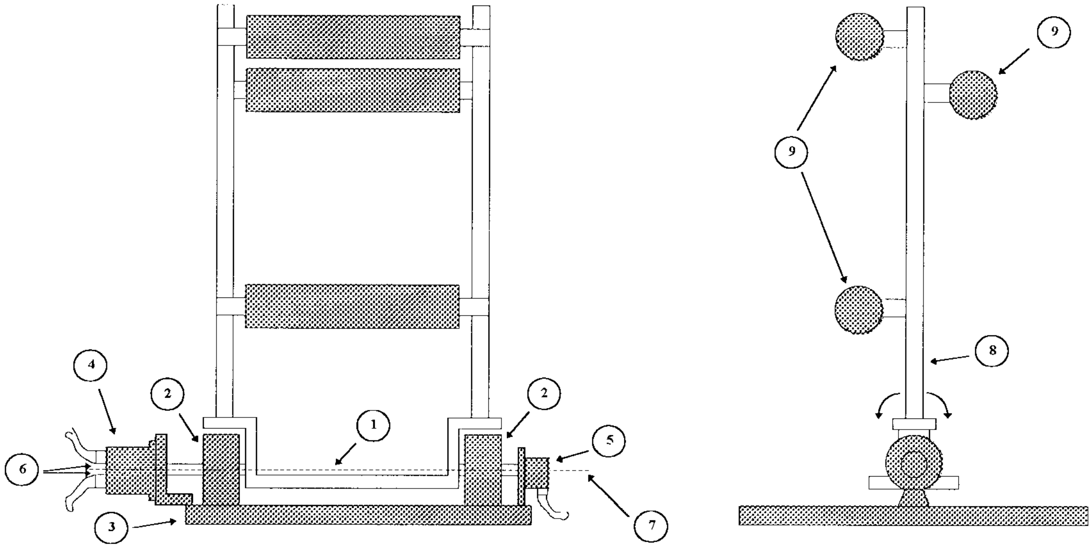


Fig. 1. Schematic presentation of the mechanical apparatus: (1) rotational platform, (2) bearings, (3) steel base, (4) hydraulic rotary valve, (5) incremental optical encoder, (6) pressure transducers, (7) artificial ankle axis, (8) vertical aluminum beams, and (9) transverse aluminum beams.

with the variations of artificially controlled lower extremities. The last experiment described in Part II was aimed to reveal whether a subject, exercising the proposed strategy, can adapt to an imperfect control of ankle stiffness.

II. Methods

A. Subjects

A healthy person aged 30 (height 170 cm, weight 80 kg) and a paraplegic aged 34 (height 185 cm, weight 90 kg), with a SCI at level T-12, participated in the experiments. Both subjects are male. The paraplegic subject was ten years postinjury and in good physical condition. In our previous work [4], we evaluated the trunk muscle abilities of both subjects in isokinetic conditions (velocity 30°/s). The performance of both subjects while seated were found to be similar (see Part I).

B. Mechanical Rotating Frame

Fig. 1 displays the frontal and lateral view of the MRF, which consists of a base and a rotational platform. The base consists of a steel plate (500 mm $\times$ 550 mm $\times$ 10 mm), bearings and a hydraulic motor. The weight of the base is 30 kg. The rotational platform (400 mm $\times$ 100 mm $\times$ 40 mm) is made of aluminum rods and plates. It provides support to the feet of the subject. It is mounted on bearings (SKF 431 700A, SKF AB, Goteborg, Sweden) which constitute an artificial ankle joint. A special bracing system, also made of aluminum alloy profiles (BOSCH, 30 mm $\times$ 30 mm), is attached to the rotational platform. Two vertical beams (1100 mm) are parallel with the legs and there are three transverse beams (450 mm), two anterior and one posterior, which maintain the subject's knees and hips in full extension (Fig. 3). The lower anterior transverse beam is below the subject's knees while the upper anterior and the posterior beams are mounted at the height of the subject's pelvis. All three transverse beams are covered with soft material. The feet of the subject are fixed to the platform by Velcro straps. The inclination of the rotating frame is measured by an optical incremental encoder (IRD 5810, PMS d.o.o, Ljubljana, Slovenia) with a resolution of 0.018°. It is not difficult for the paraplegic to get into the MRF. The posterior beam can be detached; the subject rises from his wheelchair using his arms and with the assistance of the therapist; and the therapist replaces the posterior beam, to lock the subject into the MRF. No further assistance is needed.

1) Hydraulic Actuator: The hydraulic subsystem provides the torque required for stiffness control of the artificial ankle joint. It consists from a hydraulic pump (Knapp VE 50/2-40, Knapp Mikrohydraulik Gmbh, Neutraubling, Germany), servo valve (MOOG 76-100, MOOG Gmbh, Boblingen, Germany), rotary valve actuator (ROTAC D 10, Knapp Mikrohydraulik Gmbh, Neutraubling, Germany), and two pressure transducers (VDO 3349.080.001, VDO INDUSTRIE MESSTECHNIK Gmbh, Frankfurt, Germany) (Fig. 2). The hydraulic pump provides a pressure of 50 bars to the servo valve, which controls the pressure difference applied to the rotary valve. The corner frequency of the servo valve is 40 Hz which is well above the maximal frequency of the human body movement (6 Hz) [5]. Instead of measuring the torque directly in the artificial ankle joint axis, we employed two pressure transducers to measure the pressures on both sides of the valve actuator wing. The pressure difference and the torque in the artificial ankle joint are related through the second order differential equation. However, since the corner frequency of the mass-spring system (mass and elasticity of the oil in the secondary hydraulic circuit) is around 1 kHz, we can consider the pressure difference, acting on the rotary valve, as a torque acting at the artificial ankle joint. The range of the pressure transducers varies between 0 and 60 bars with the accuracy of 0.5%, equivalent to 0.3 Nm of torque.

2) Stiffness Control: The MRF is controlled by software (C language) running on a personal computer (Intel 486, 66 MHz IBM compatible PC) equipped with data acquisition units (Burr–Brown PCI-20001C). These perform A/D and D/A

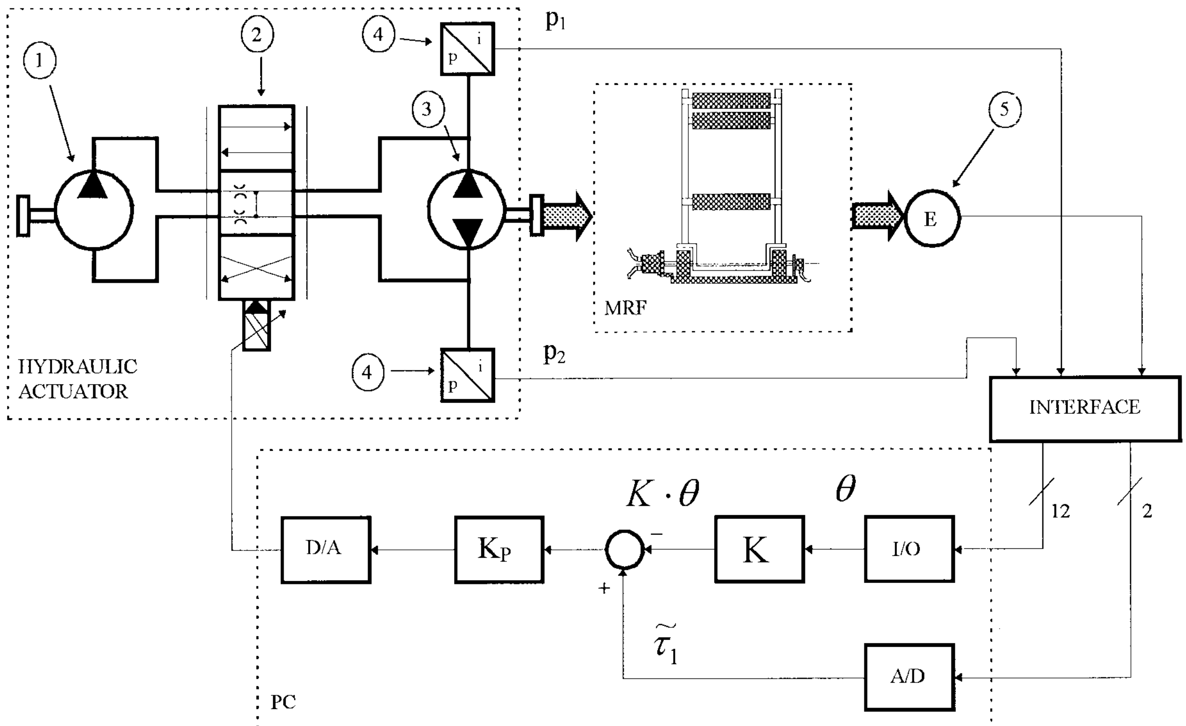


Fig. 2. Schematic presentation of a complete MRF device. Hydraulic subsystem consists from (1) hydraulic pump, (2) servo valve, (3) rotary valve, (4) pressure transducers, and (5) encoder. The encoder and pressure transducer signals (pressures $p1$ and $p2$) are interfaced to the software implemented control algorithm running on a PC. θ is a sway angle in the artificial ankle joint, $\tilde{\tau}_1$ is the estimated torque acting around the artificial ankle joint produced by the hydraulic actuator. K represents the desired stiffness value and Kp is the gain of a proportional controller.

conversion and I/O interfacing. A special hardware interface transforms the signals of the incremental encoder and pressure transducers into the form appropriate for data acquisition. The interface also determines the absolute angle (12 bits) of the artificial ankle joint. The control subroutine calculates the reference torque, according to the desired ankle joint stiffness, and compares it to the actual torque. The error signal passes through a simple P controller and, after D/A conversion, drives the servo valve (Fig. 2). The sampling frequency of the control loop is 300 Hz. The simple proportional controller was found to be sufficiently robust, reliable and to provide adequate tracking of the reference torque. Apart from maintaining the required stiffness, the software also allows perturbations to be applied to the artificial ankle joint, and simulation of ankle muscle fatigue by exponentially decreasing the stiffness constant K with time.

3) Influence of MRF on the Balancing Abilities of the Standing Subject The rotational part of MRF weights 18 kg, the center of gravity is 0.2 m above the axis of the artificial ankle joint and the moment of inertia around the joint is 3 kgm^2. In comparison to the corresponding parameters of the lower part of the body (mass 40 kg, center of gravity 0.67 m above the ankle joint and the moment of inertia around the ankle joint 18 kgm^2 [5]), the rotating frame does not add much to the inertia of the lower link of the double inverted pendulum. In Part I, it was shown that the coupling index is not sensitive to the inertia parameters of the links so the addition of the MRF should not be significant. The friction present in the artificial ankle joint was found to be 2.5 Nm. Trnkoczy *et al.* [6] identified a Coulomb friction moment in a paralyzed ankle joint to be 1.2 Nm, so the friction from two real ankles is very nearly equal to the friction in the MRF.

4) Safety Features of the MRF Device: The range of the platform rotation is limited mechanically to ±20° (0° is vertical). The subject wears a full-body harness, loosely coupled by ropes to the ceiling. The ropes are for safety, to limit the range of positions of the trunk if the subject falls. Fig. 3 shows an intact subject standing in MRF, wearing the harness: the safety ropes can be clearly seen. Before the experiments, we demonstrated to each subject how the harness and the safety ropes prevented falling. When they realized that, even if they fell over, nothing harmful would happen, they relaxed and were ready to focus on the task of balancing. The interface hardware includes also a panic button, housed in a separate chassis, which can be pressed by the therapist in the event of system malfunctioning. Another safety precaution is inherent in the procedure of getting the subject in and out of the MRF. There is a space between the steel base and the platform (see Fig. 3). Before the subject enters the frame, this space is filled with a close-fitting wooden plate, which reduces the inclination range of the frame to approximately ±1°. After the subject is secured in the bracing frame, the device is tested while the wooden plate is still in place. Only after normal performance of the system has been confirmed is the wooden plate removed. The reverse procedure is employed when releasing the subject from the MRF.

C. Sensory Feedback Implementation

The microcontroller evaluation board (Motorola MC68HC16) was used to implement the auditory feedback.

Fig. 3. An intact subject standing in the MRF. The restraining aluminum beams, standing platform and safety ropes are clearly shown. The hydraulic actuator and incremental encoder are also visible.

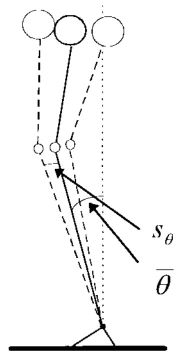
The artificial ankle angle magnitude and direction signal was transmitted via a serial data link from the PC controlling the OPTOTRAK[1] system every 20 ms. The inclination (0–10°) was transformed exponentially into sinusoidal tones of frequency ranging from 100 to 1000 Hz. This frequency range was used since it encompasses a large set of just-noticeable frequency differences and provides a pleasant sensation [7]. The amplitude of the auditory signal was adjusted to a comfortable level. When the lower link was inclined in the forward direction (positive ankle angle) the amplitude of the sound was set to 100% of the adjustment level, while in the case of backward inclination, it was 75%. In this way the subject was able to distinguish between the two postures. Two loudspeakers were used, one placed in front and the other behind the subject, at 2 m distance and at the height of his head. We selected auditory feedback mainly because it avoided encumbering the subjects with additional wires and stimulators, which would have been necessary for electrocutaneous or vibrotactile feedback. Unlike the feedback setup in [8], we did not implement a "dead space" in a sensory signal, for small deviations from the upright posture. The transition between the two different amplitude levels, when ankle angle passed zero, marked the exact upright position.

[1] OPTOTRAK is a registered trademark of Northern Digital, Inc.

D. Experimental Conditions

The subject was positioned in the MRF as shown in Fig. 3. The subject's ankle axis was 2 cm above the artificial ankle joint, while the pelvis and the feet were positioned in such a way that the lumbosacral joint axis and the ankle joint axis of the subject lay in the plane of the midlines of the vertical beams of the bracing system. The OPTOTRAK[1] optical system was used to measure the movement of the double inverted pendulum. Three infrared active markers were positioned on the MRF and the upper trunk of the subject. The first marker was attached to the shaft encoder on the artificial ankle joint axis; the second was attached to the vertical beam of the bracing system at the height of a subject's lumbosacral joint (L5–S1); while the third was placed on the midline of the trunk, half way between the iliac crest and the shoulder. The values of both angles θ and ψ, defined in Fig. 5, were calculated from the markers' positions. The MRF base was firmly fixed to the force plate (AMTI OR6-5-1) which measured the reaction forces and torques during the experiments. The marker positions and the reaction forces and torques were sampled at 50 Hz and saved for off-line analysis. The torque τ_2, defined in Fig. 5, acting at the lumbosacral joint was calculated from the reaction forces and torques and the kinematic variables of the lower body [9]. The inertia properties of each subject were estimated from [5]. The first and most important question to be answered in this investigation was whether both subjects are capable of proposed underactuated balancing when constrained by the MRF. There was no doubt that with high stiffness in the artificial ankle joint (well above $K_v = 11.2$ Nm/°, as introduced in the Part I) the subject would be able to balance. It was, therefore, our aim to investigate and evaluate the performance of both subjects at lower values of stiffness. This experiment was entitled "quiet standing." Another important question was whether the subjects were able to recover from disturbances, imposed in the artificial ankle joint, and which strategies are used for recovery. We called these experiments "perturbed standing." The third experiment, "blindfold standing," was aimed to reveal the importance of cognitive sensory feedback, when vision was disabled. Finally, we were interested in the abilities of the subjects to balance when the ankle stiffness was varied, to imitate ankle muscle fatigue, as posture switching would be essential for continued standing. This last experiment was entitled "fatigue simulation." In all experiments the subjects were allowed to maintain balance only by trunk movements around the lumbosacral joint.

1) Quiet Standing: The two subjects followed the same protocol. We expected that learning would be important for both subjects, so we arbitrarily selected a training period of five consecutive days. At the beginning of each training day, the subject stood in the MRF at a high level of stiffness (10 Nm/°) which was comfortable and stable. This initial stand lasted for approximately 10 min and enabled the subject to become familiar with the task and experimental environment. After this initial period, the stiffness in the MRF ankle joint was reduced to 8 Nm/° and a set of five standing trials



$$Posture = \left|-K \cdot \overline{\theta}\right|$$
$$Postural\_activity = K \cdot s_\theta$$

Fig. 4. Illustration of the standing evaluation variables, Posture and Postural activity, which describe the quality of a particular successful trial. $\overline{\theta}$ denotes the mean value of the ankle angle during a particular successful trial. s_θ represents the standard deviation of the ankle angle for a particular successful trial. K denotes a selected ankle stiffness.

followed, during which the subject was told to maintain an upright posture, as far as possible. Since it was unlikely that the exact upright posture could be assumed and maintained, he was also instructed to try to achieve a forward posture (lower body inclined in anterior direction, see Part I), close to upright. The backward posture (lower body inclined in posterior direction) was less desired, since the forward posture was expected to be more robust following disturbance (see Part I, Section III). If the duration of balancing exceeded 20 s, the trial was considered successful. After each set of five trials, the stiffness was reduced by 1 Nm/°. This was continued until the subject was unable to accomplish at least one successful trial out of five at that stiffness, or until the paraplegic subject was too tired to continue. The first day was regarded as introductory and no data was recorded. On day 2, the experiments of the first day were repeated and kinematic and dynamic data were collected. During days 3 to 5, cognitive feedback was provided. The arms of both subjects were folded, on the chest for the intact subject, and on the back for the paraplegic.

The accuracy of the lumbosacral angle measurement is likely to be poor, due to the movement of the trunk marker relative to the body, and simply because the upper trunk is not rigid. However, the measurements of the torque and angle at the artificial ankle joint were accurate. We, therefore, decided to evaluate quiet standing on the basis of the ankle torque measurements. We calculated the mean value and standard deviation of the ankle torque for each successful trial (Fig. 5). We named the mean ankle torque a “Posture” and the standard deviation of the ankle torque a “postural activity.” These two characteristics, illustrated in Fig. 4, were used for the evaluation of the standing performance. We then calculated the means and standard deviations of posture and postural activity for all successful trials at each value of ankle stiffness. The absolute values of the Posture for a particular trial, were used in the calculation, since it can be either positive (backward posture) or negative (forward posture).

2) Perturbed Standing: After the completion of the five days lasting quiet standing experiment, the perturbed standing experiments were conducted with the stiffness set to 8 Nm/°. The subject was instructed to assume a near-upright posture and stand as still as possible. With a stroke on the keyboard, the operator induced the perturbation torque impulse (50 Nm, 100 ms) in either an anterior or posterior direction. He sat behind the subject so this action was invisible to the subject. Each subject was exposed to twenty trials of anterior and posterior perturbations. In the first ten trials (five anterior perturbations and five posterior perturbations in random order), cognitive feedback was used, while in the remaining ten (also five anterior perturbations and five posterior perturbations in random order), it was not.

In perturbed standing, the electromyograms (EMG’s) of the abdominal and paraspinal trunk muscle groups were monitored. A similar measurement setup was described by Horak and Nashner [10]. The muscle activities were recorded by surface electrodes (Axelgaard, diameter 2.5 cm). The electrodes monitoring the paraspinal muscles activity were placed at the level of the iliac crest (L4–5, primarily erector spinae) while the electrodes monitoring the abdominal activity were placed at the umbilical level (primarily rectus abdominis). The interelectrode distance was 3 cm in both cases. Precision differential amplifiers (frequency band 50–5000 Hz, gain 5000) were used to preprocess the signals. The sampling rate was 500 Hz. Both EMG signals were full-wave rectified and low-pass filtered in both directions, thus preserving the phase content of the signals (fifth-order discrete Butterworth filter, cutoff frequency 10 Hz, implemented by MATLAB[2]). The voltages of both EMG signals were in approximate proportion to the level of muscle activation but no attempt was made to calibrate the EMG signals on an absolute scale.

In this experiment, we were interested in the latencies of the postural loop during recovery from disturbances as well as the strategy that each subject used for recovery.

3) Blindfold Standing: In this experiment, which was conducted a day after the Perturbed standing experiment, we investigated the importance of cognitive feedback in quiet standing in both subjects when standing blindfolded. The stiffness value was set to 8 Nm/°. Both subjects performed five trials with the cognitive feedback and five trials without it.

4) Fatigue Simulation: In Part I, we hypothesized that the subjects should be able to switch posture, when the ankle torque, produced by the ankle dorsiflexors when standing in a backward posture, or the plantarflexors when standing in a forward posture, decreased due to muscle fatigue. In these experiments, we simulated fatigue in the artificial ankle joint by exponentially decreasing the stiffness from the initial preset value of 8 Nm/°, halving in 20 s while in either the forward or backward posture. The stiffness was automatically reset to the initial value after the subject switched posture. The ability of the intact subject was investigated in five standing trials with cognitive feedback and five without.

[2] MATLAB is a registered trademark of The Math Works, Inc., Natick, MA.

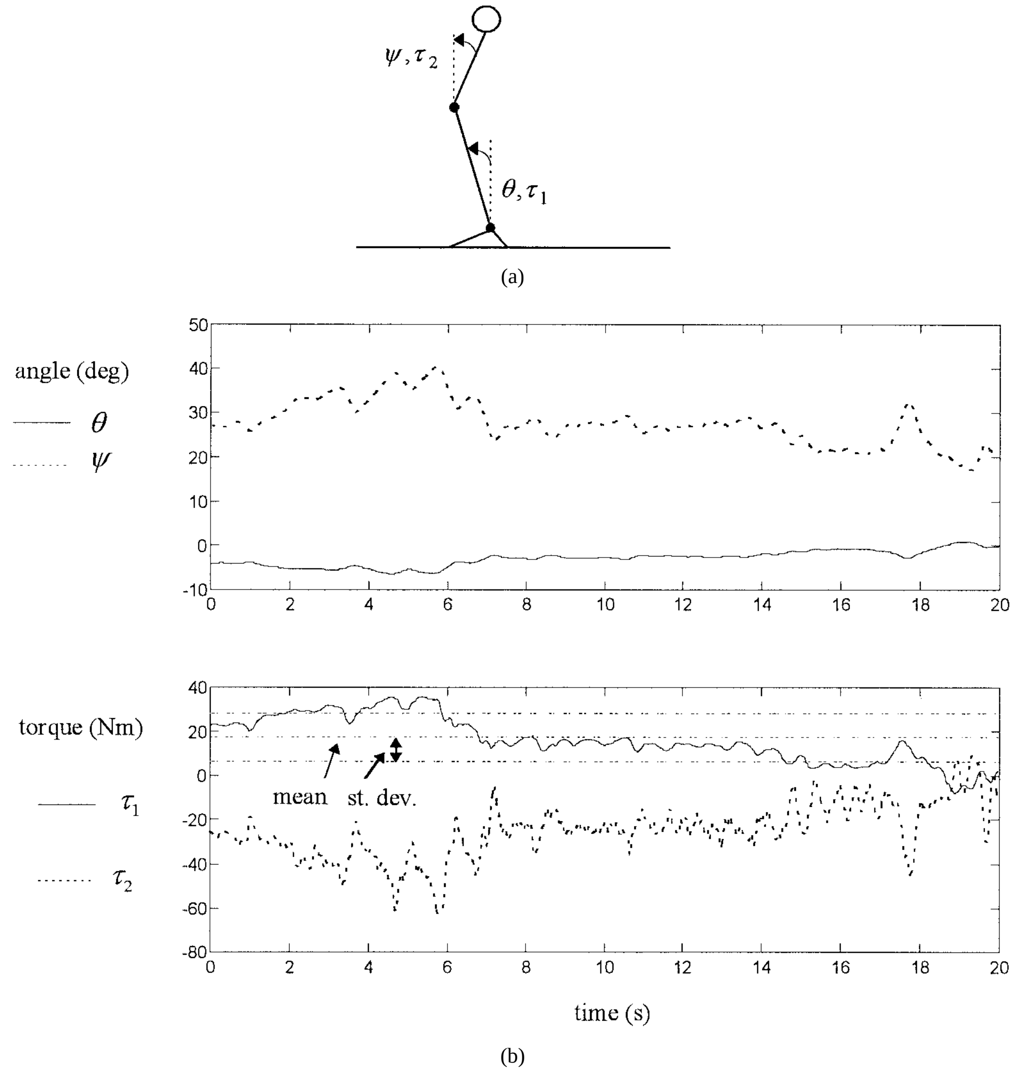


Fig. 5. (a) Definition of ankle and lumbosacral angles and torques and (b) a representative record of quiet standing by the paraplegic subject. The ankle stiffness was set to 6 Nm/° and cognitive feedback was applied. The subject maintained the backward posture throughout the trial.

III. Results

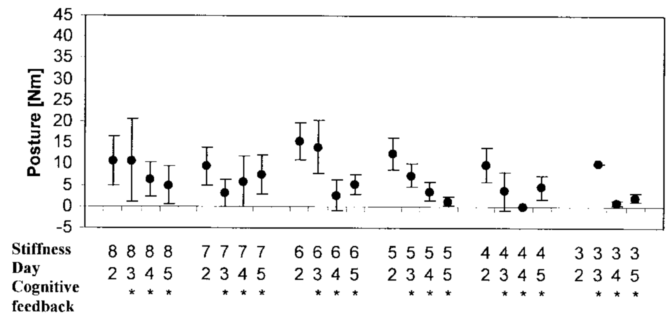
A. Quiet Standing

Fig. 5 shows a representative successful trial of the paraplegic when the artificial ankle stiffness was set to 6 Nm/° and cognitive feedback was provided. The angle θ and the torque τ_1 were measured in the artificial ankle joint while ψ and τ_2 belong to the lumbosacral joint. During the first 5 s of the trial the backward inclination of the lower body was quite high (−5°). Between 5 and 7 s, the subject voluntarily assumed a more upright posture. Near the end of the trial, he decreased the inclination even further.

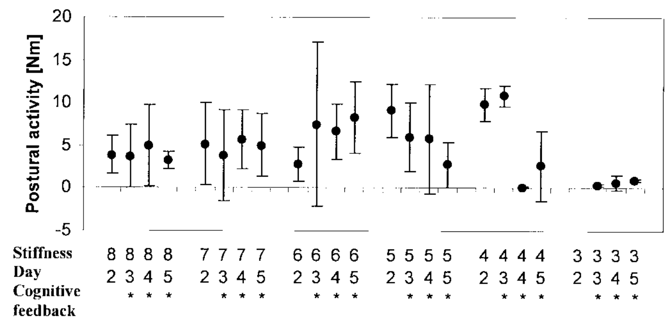
The evaluation results for the intact subject are shown in Fig. 6. For an illustration of the evaluation procedure, let us consider the standing performance on the second day with the ankle stiffness set to 8 Nm/°. From Fig. 6(c), showing the number of successful trials, we see that the intact subject was successful in all five trials and that he always assumed the backward posture. From Fig. 6(a), we see that the mean value of the Posture for five trials was slightly over 10 Nm and its standard deviation was around 6 Nm. The mean value of the Postural activity is shown in Fig. 6(b) and was slightly below 4 Nm with the standard deviation around 2 Nm.

By comparing the performance of the intact subject at the stiffness level of 8 Nm/°, over these last four days of the experiment, we observe that the mean value of the Posture decreased, indicating that the subject learned how to attain a more upright posture. The mean value of the Postural activity remained almost the same, indicating a constant level of balancing effort. For the stiffness value of 7 Nm/°, the mean value of Posture on day 3 and day 4 were lower than on day 2 and day 5. The mean values of Postural activity were comparable throughout this period and slightly higher than at 8 Nm/°. Here, we have to say that the subject balanced with little effort when the stiffness was set to 8 and 7 Nm/°: while standing he could talk to the people in the laboratory and move his arms without becoming unstable, which indicates the low level of effort he had to put into his task.

In trials where the stiffness value was set below 7 Nm/°, he had to pay more attention to the balancing task. From Fig. 6(a)



(a)



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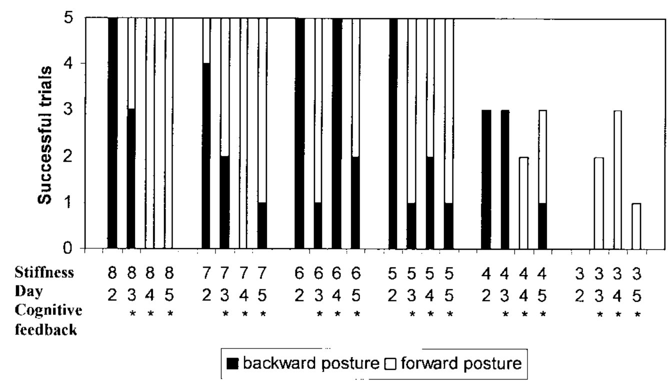
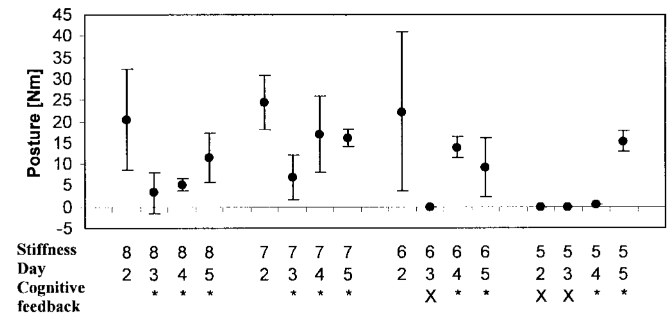
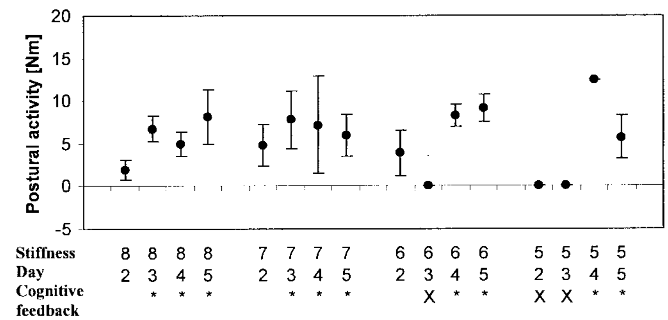


(c)

Fig. 6. The results of quiet standing for the intact subject for four successive days: (a) means and standard deviations of the Posture for successful trials at different stiffness levels, (b) means and standard deviations of the Postural activity for successful trials at different stiffness levels, and (c) number of successful trials at different stiffness levels. Last three days when a cognitive feedback was delivered are denoted by $*$.



(a)



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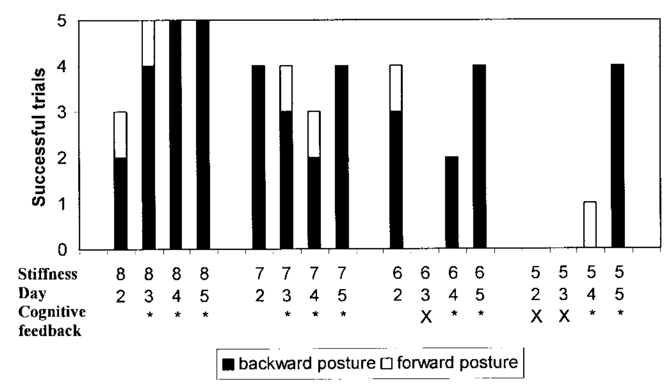

(c)

Fig. 7. The results of quiet standing as assessed in the paraplegic subject for four successive days: (a) means and standard deviations of the posture for successful trials at different stiffness levels, (b) means and standard deviations of the postural activity for successful trials at different stiffness levels, and (c) number of successful trials at different stiffness levels. Last three days, when a cognitive feedback was delivered, are denoted by $*$. X denotes sets of trials, at particular stiffness level and on a particular day, which were not done due to the subject's fatigue.

it can be seen that the mean value of the Posture on day 2, at the stiffness value of 6 Nm/°, was higher than at 8 and 7 Nm/°. During days 2–5, the subject managed to decrease the Posture, but the Postural activity increased. A similar observation can be made also for the Posture when the stiffness was 5 Nm/°, however the Postural activity decreased.

The subject had to give his full attention to the task when standing at 4 and 3 Nm/°: in contrast to trials at higher stiffness values, not all trials were successful. Over the period from day 2 to day 5, no improvement in the fraction of successful trials was observed. When standing at the stiffness value of 3 Nm/°, there was little Postural activity. The subject was aware that balancing was difficult so, after he was initially placed in a stable posture, he did not dare to breathe in order to avoid disturbances which could cause a fall.

From Fig. 6(c) it can be seen that, on day 2, the subject mostly utilized the backward posture but changed so that, on day 5, he usually assumed the forward posture.

Fig. 7 presents the results of quiet standing by the paraplegic subject. Performance similar to the intact subject was found when the ankle stiffness was 8 Nm/°. The main difference between the two subjects was the posture: the paraplegic preferred the backward posture. The reason for this was contractures of his iliopsoas muscles. Although he was able to attain the forward posture, balancing was not then successful since the lumbosacral joint was near the limit of motion. He was unable to perform all the balancing trials at lower values of ankle stiffness. A serious obstacle to his balancing in the

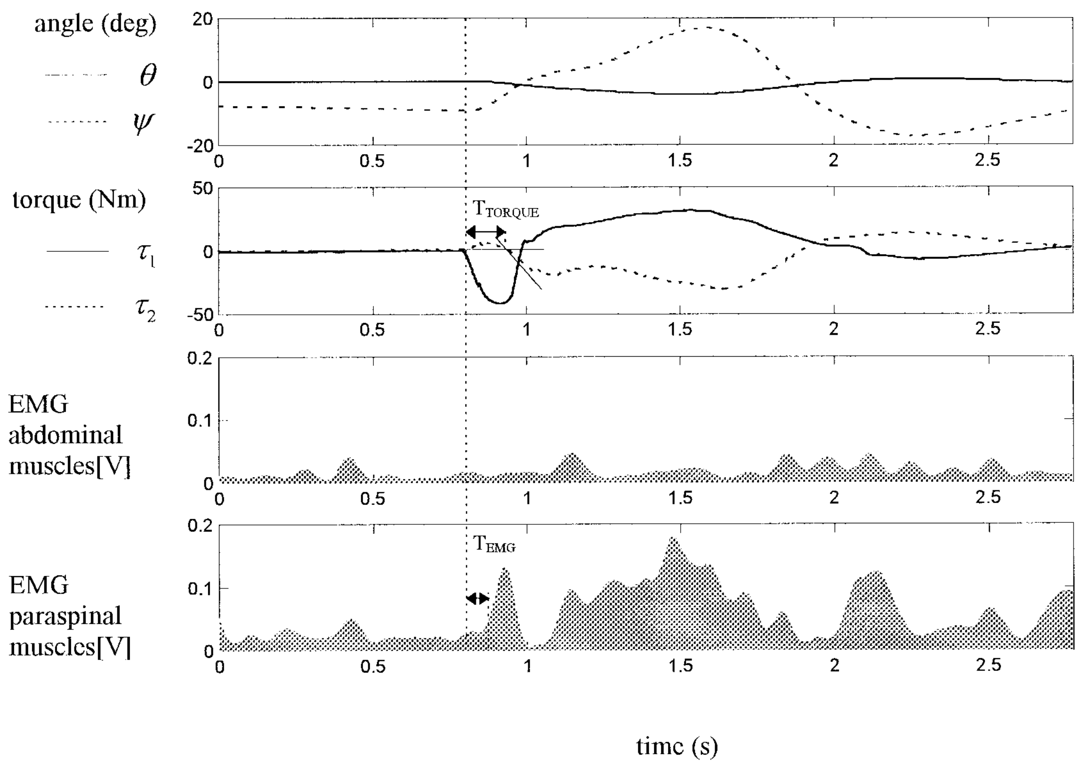

Fig. 8. Disturbance recovery by the intact subject. Both angles, torques, and abdominal and paraspinal muscles' EMG responses are shown. The latencies T_{TORQUE} and T_{EMG} in the lumbosacral joint are also depicted. The vertical dashed line denotes time when the disturbance commenced.

MRF was fatigue of the upper body, which was pronounced on day 1 and day 2 of the experiment. The marks X, in Fig. 7, denote the trials that he omitted due to fatigue or pain in the lumbosacral joint. However, after he had completed all the experiments (7 days), he was able to balance in the MRF at 8 Nm/° for half an hour with no arm support and without becoming tired.

The range of ankle stiffness, from 8–3 Nm/°, was tested because we wanted to investigate the limitations of under-actuated balancing. However, the performance of the intact subject when standing at high stiffness values ranging from 30 to 10 Nm/° was also assessed. The results for the posture and postural activity were very similar to the results presented in Fig. 6 for stiffness level of 8 Nm/°.

Both subjects found the auditory feedback very useful. When the feedback was not delivered to them they were not aware of the exact position of their lower body. After the completion of the quiet standing experiment both subjects performed a few trials without auditory feedback at the ankle stiffness of 8 Nm/°. Although they had no difficulties with balancing, they preferred to have the auditory signal. They stated that, in the presence of the cognitive feedback, they felt more confident of their performance.

B. Perturbed Standing

When applying perturbations to the artificial ankle joint, we tried to induce the disturbance only after the angles and torques in both joints remained constant for at least one second, thus assuring that the observed response was the result of the induced perturbation only. We managed to achieve these conditions for the intact subject, but we were not always successful with the paraplegic.

Fig. 8. shows a representative sample of the intact subject's response to a sudden perturbation, acting in the posterior direction, and applied at approximately 0.8 s. Immediately after the onset of the disturbance, we observe a small change in the calculated (not measured) torque in the lumbosacral joint. Since there was no coincident EMG activity, this early rise can only be due to the passive properties of the joint, unless it is a measurement artifact due to inaccurate estimation of the inertia properties (masses, centers of masses and moments of inertia of the body and the MRF) used in the calculation. After approximately 100 ms, there is a rise in the EMG of the paraspinalis muscles and, after approximately another 50 ms, the lumbosacral torque started to increase. The strategy for recovery can be recognized from the time course of both torques and the lumbosacral angle (Fig. 8). The subject first stabilized himself in a new equilibrium posture. Once he was stabilized, he voluntarily initiated activity to regain an upright posture again. These experimental results resemble the simulated response to disturbance [Part I; Fig. 8(b)]: similar strategies appear in both cases.

In the first ten trials, when cognitive feedback was provided to the standing subject, we found no statistically significant difference between the latencies of the response resulting from the anterior and posterior perturbations. We, therefore, calculated the mean and the standard deviation of latencies for all ten trials irrespective to the direction of the perturbation. In Table I the latencies in the EMG (T_{EMG}) and lumbosacral torque (T_{TORQUE}) are shown for the intact subject with and

TABLE I
LATENCIES FOR THE INTACT AND PARAPLEGIC SUBJECT

	INTACT SUBJECT		PARAPLEGIC SUBJECT	
	FEEDBACK	NO FEEDBACK	FEEDBACK	NO FEEDBACK
T_{EMG}	105 ± 14 ms	101 ± 13 ms	-	-
T_{TORQUE}	156 ± 5 ms	153 ± 5 ms	142 ± 8 ms	146 ± 5 ms
SUCCESSFUL TRIALS	10	10	10	6

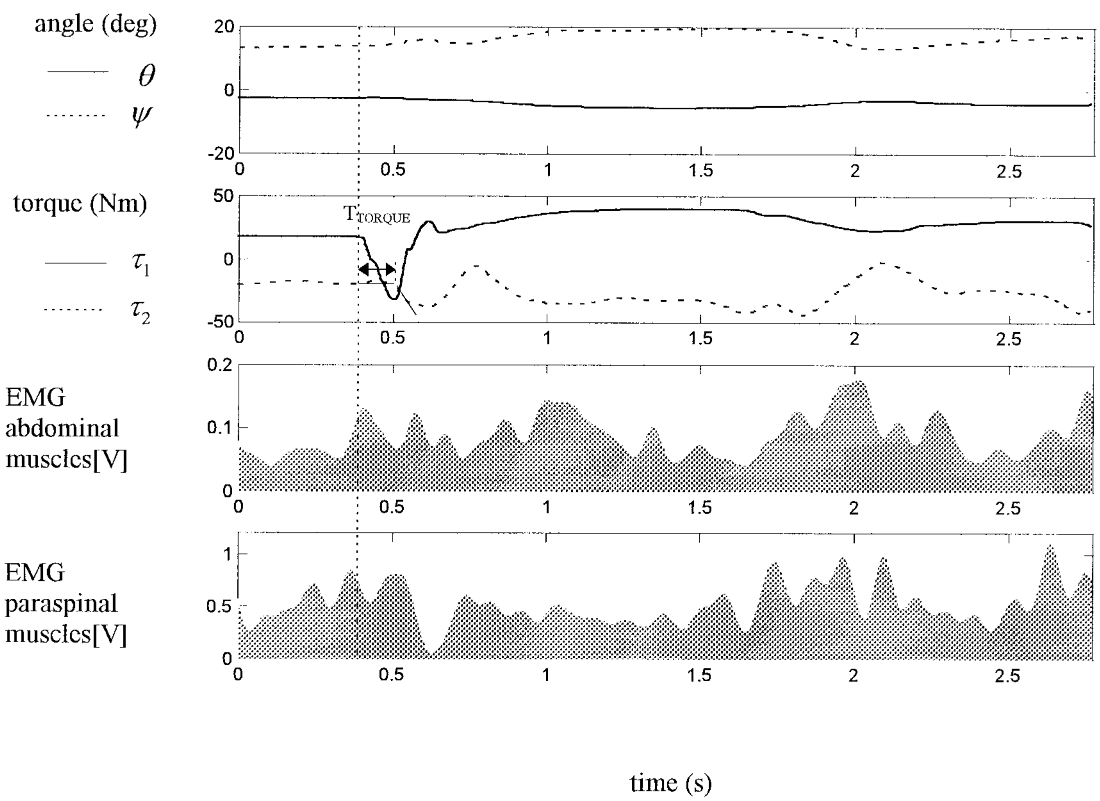

Fig. 9. Disturbance recovery while the paraplegic subject is standing. Both angles, torques and abdominal and paraspinalis EMG signals are shown. The latency T_{TORQUE} in the lumbosacral joint is also depicted. The vertical dashed line denotes time when the disturbance commenced.

without cognitive feedback. The differences of the mean latencies, both for EMG and lumbosacral torque, were statistically insignificant ($\alpha = 0.05$, one-tailed *t*-test), suggesting that the auditory feedback has no influence on the latency of the posture control loop.

Typical disturbance responses for the paraplegic are displayed in Fig. 9. We see that he was not in an upright posture prior to the disturbance. His backward posture causes EMG activity which made it impossible to estimate the latencies of the EMG response. We could, however, still measure the latencies in the lumbosacral torque (T_{TORQUE}). The trajectories of the lumbosacral angle and torque, during the first second of his response, are similar to the response of the intact subject, indicating that they both use a similar strategy for disturbance recovery. However, the paraplegic exhibits a longer oscillatory response when regaining an upright posture after perturbation.

Table I also presents the paraplegic's torque latencies. As in the intact subject, the difference between standing with and without cognitive feedback were statistically insignificant ($\alpha = 0.05$, one-tailed *t*-test). However, when trying to balance without cognitive feedback, the paraplegic failed four times to recover from the disturbance. The reason for these failures appears to be poor initial posture, which in unsuccessful trials, was always far from upright. Without cognitive feedback, the intact subject also sometimes stood far from upright but, after the disturbance, he was able to recover due to the greater range of motion in his lumbosacral joint.

C. Blindfold Standing

Both subjects were successful in all five trials with cognitive feedback but invariably failed without it.

D. Fatigue Simulation

Fig. 10 shows a representative record of a successful 60-s standing trial. As the ankle torque decreased, the subject voluntarily changed the backward to the forward posture in the tenth second. At the instant when the switching occurred, an oscillatory transient can be seen in both torques. As the stiffness in the ankle decreases exponentially, it is crucial that the subject does not wait too long before initiating the switching maneuver. Five clear switchings can be seen during

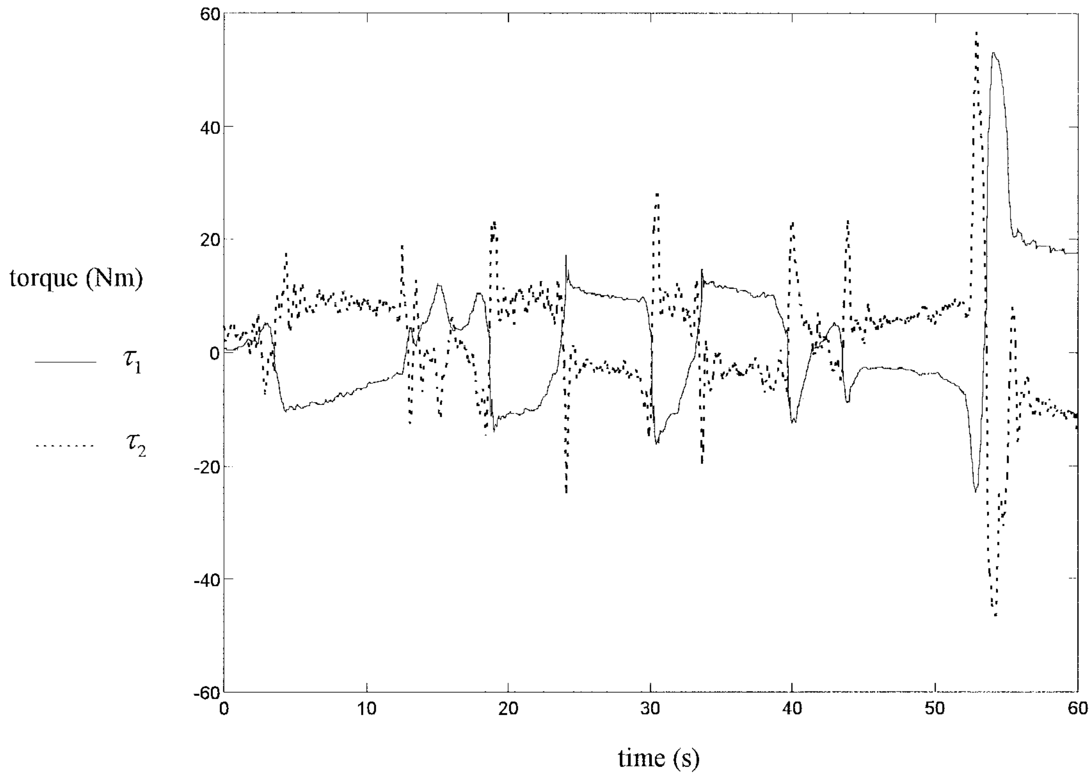

Fig. 10. A representative time course of ankle and lumbosacral joint torques during the fatigue simulation experiment on the intact subject using cognitive feedback.

the period from 20 to 40 s. The maneuver which occurred around the 50th second was initiated later, at lower stiffness than the previous maneuvers, and the consequent difficulty in balancing can clearly be seen.

As in previous experiments, all five trials with cognitive feedback were successful, while without this feedback, none of the attempts was successful. The cause of failure was the subject's lack of information about the instantaneous ankle joint position causing him to start the switching maneuver too late.

This experiment was not conducted with the paraplegic subject because his iliopsoas contractures gave only restricted movement in the forward posture.

IV. Discussion

A. Experiments

The feasibility of the underactuated control strategy for arm-free standing, in a paraplegic and an intact subject, constrained in the mechanical rotating frame (MRF) has been demonstrated in our experimental work.

The "quiet standing" experiment confirmed that the residual sensory and motor functions as well as the CNS of the paraplegic subject are sufficient for maintaining arm-free standing in the presence of adequate ankle stiffness. Considering the results of the intact subject, presented in Fig. 6, when standing with a stiffness in the range from 8 to 5 Nm/°, one can see that a mean value and standard deviation of Posture decreased over the four days of training, which indicates that the subject developed the skills for balancing. In contrast, the paraplegic, while balancing, had to cope with contractures and, during the first two days, also with pain in the lumbosacral joint. We think this may be the reason why similar improvement was not seen in Fig. 7. However, in this study, both subjects could balance even without cognitive feedback (Day 2) and the paraplegic's performance is quite robust when standing with 8 Nm/°. During the last three days, when cognitive feedback was provided, the posture selected by both subjects was consistent. The main reason for this change may be the cognitive feedback, which provided both subjects with postural information.

The "perturbed standing" experiment revealed that, when perturbed, both subjects used a similar strategy for recovery. There were no significant differences in the latencies of their responses, suggesting that similar sensory systems, presumably the intact vestibular organ, the visual system, and upper body proprioception, are involved in postural control (see also discussion about the MRF).

The "blindfold standing" experiment confirmed our assumption that the visual system provides a reference feedback signal which is used by the subject. It is also clear that a cognitive feedback signal can adequately replace vision.

In the "fatigue simulation" experiment, in which posture switching was the only way to maintain balance, the subject demonstrated that he could make the transition and could usually judge when to do so. For this experiment to be successful, we showed that cognitive feedback is essential.

Our subjects were not allowed to use their arms for balancing. We expect that dynamic forces, resulting from arm movements around the shoulders, would enable the lumbosacral and ankle torques to be decreased. By moving the arms, balance would probably be improved Nevertheless, it is our belief that the use of arms should be preserved for functional manipulation rather than for balancing.

B. Mechanical Rotating Frame Issues

Nashner [11] designed a special standing platform with two degree of freedom, one translational and one rotational, which enabled him to investigate the role and dynamic properties of the vestibular system in perturbed human standing. Body sway was induced by translational movement. The sensory feedback, arising from the ankle joint deflection, were removed by rotating the platform base in order to maintain a constant ankle angle. He stated that this removed reflex feedback and eliminated cues from the joint and deep pressure receptors. If we consider the position of the legs of a subject in the MRF, we realize that the ankle angles do not change, as for Nashner's standing platform. This means that an intact subject, standing in MRF, not only loses the motor abilities of his lower extremities, but presumably also sensory information from the ankle joints. However, he may still receive sensory information from soles of his feet and exteroceptive information from contact with the transverse bars of the MRF. Even though similar responses were observed in both subjects, during the "perturbed standing" experiment, which leads to a hypothesis that only the vestibular and visual systems, and proprioception of the upper body are included in the postural loop of a paraplegic (since the lower-body proprioception is clearly of no use in paraplegic), we do not know whether the lower-body proprioceptors of an intact subject are also included in his control of posture. However, Allum and Honegger [12] showed experimentally that the balance-correction strategies of normal subjects derive from vestibular and visual information but are independent of the local sensory input gathered from the lower limbs. This is extremely important for future investigations of underactuated arm-free standing because it enables us to conduct future exploratory work with intact subjects.

An important requirement, that was successfully fulfilled by the MRF, was the safety of a standing subject. This was not only important for preventing injury but, we think, also had the important psychological effect that both subjects lost their initial sense of insecurity.

From these preliminary experiences with the MRF, we think that the device could be a useful therapy for complete paraplegics. The stiffness in the artificial ankle joint can be set, so that any paraplegic can balance, however poor are his or her strength and voluntary control of the trunk muscles. By everyday training in the device, the trunk muscles that are under voluntary control will restrenghthen, and the range of motion of the lumbosacral joint will increase. In this regard, the device offers important advantages over passive standing frames, currently used in rehabilitation centers and patients' homes. It provides balance training, which is required for standing and walking assisted by either mechanical braces or functional electrical stimulation. Besides having a tremendous psychological impact, its use will probably also bring medical benefits [1], [13].

C. Experimental Findings in the Light of Theoretical Predictions

The experiments described in Part II confirmed our assumption, made in Part I, that this method resolves the problem of synchronizing the voluntary activity of the upper body with the artificial control of the paralyzed lower limbs. Since the task of the ankle controller is only to regulate the stiffness, and not to position the ankles, the problem of determining the reference angle, when trying to adopt a new posture, does not exist. The action of the artificial controller is "passive"; the actions of the upper body control the posture.

We proved experimentally that no artificial sensory feedback of ankle angle, is necessary for balance, when utilizing the underactuated control strategy. The preserved natural vestibular and visual sensory systems together with proprioception of the upper body, neck, and head are sufficient, as suggested in Part I. However, balance, in presence of disturbances and uncertainties, is more robust with cognitive auditory feedback.

Comparing the time courses of lumbosacral angles and torques in the "perturbed standing" experiment with the simulation study (Part I, Fig. 8), we see similar time delay, shape and peak magnitude during the first 300 ms of the response, indicating that, at least in the first approximation, the closed-loop model is valid. The experimental responses after 300 ms following the onset of the disturbance, differ from the simulated responses, primarily due to different postural activity, seen in both subjects. Experimental responses from both subjects also confirmed that the strategy for recovery was that predicted in Part I, Fig. 8(b).

Results from the theoretical analysis of feasible disturbance space showed that the optimal value of ankle stiffness should be around 10 Nm/degree when considering results for all three groups of posture (forward, backward, and upright). It is therefore very encouraging that the "quiet standing" experiment showed robust balancing, requiring little effort from both subjects, when the ankle stiffness was only 8 Nm/°.

D. Implementation of the Proposed Arm-Free Standing

There are two possible methods for providing the necessary stiffness at the paraplegic's ankles.

- The first is passive mechanical braces or elastic rods with proper stiffness, mounted in both shoes. These shoes should be designed to have the required stiffness for ankle flexion and extension, but also to prevent eversion and inversion so that lateral stability is assured. Since the passive stiffness in the ankle joint is often already increased by spasticity and contractures [14], [15], the extra flexion and extension stiffness from the orthosis may be rather small. The knee and hip joints can be locked in extended positions by open-loop FES, perhaps with posture switching, to cyclically engage and disengage the hip and knee extensor muscles, as proposed by Kralj [16]. Jaeger [17] pointed out that FES systems for providing unsupported standing to paraplegics must be sufficiently simple to be implemented in clinical use. A system for standing without arm support, which uses mechanical springs at the ankles, would be appropriate.
- The second approach would use closed-loop control of electrical stimulation, delivered to the agonist and antagonist muscle groups of the ankles. Stabilization of the other joints can be accomplished in the same way as in the first approach.

Promising work regarding the closed-loop control of the ankle plantarflexor muscles was recently accomplished by the group in London [18], [19]. They proposed and implemented the control of unsupported standing of the intact and paraplegic person. They braced the whole body of a standing subject and controlled it as a single inverted pendulum by an LQG controller with three nested loops. They demonstrated the robustness and good tracking performance of the controller, however, they stated that fatigue of the electrically stimulated muscles and spasticity prevented prolonged standing. By combining their ankle controller and our underactuated double inverted pendulum we might obtain a rehabilitative system which enables prolonged standing, since we have demonstrated the abilities of a paraplegic subject to switch posture and recover from disturbances that may be caused by spasms.

The same authors [18], [19] have also pointed out the need for accurate ankle position and torque sensors, as well as identification of the stimulated muscle properties. Since our control scheme can cope with ankle torque variations, closed-loop control of the ankle does not need to be optimal. Instead of using the LQG theory for the controller parameters synthesis, we could use simple PID controllers. The effects of nonoptimal control in the ankles could be compensated by voluntary upper body activity.

The research group in Aalborg has demonstrated the feasibility of extracting reasonably accurate information on the center of pressure (COP) position from the natural sensors in the sole of the feet [20]. In future we could utilize this sensory information in the control of arm-free standing. As we have demonstrated in the "fatigue simulation" experiment, the artificial control and thus also its sensory input does not need to be absolutely accurate in our method.

We will continue our experiments using mechanical springs at the ankles and the open-loop FES of the knee and hip extensors. Without the inertia of the MRF, the effort required of the lumbosacral joint should be reduced.

V. Conclusions

In this study we have experimentally demonstrated the feasibility of unsupported paraplegic standing, utilizing the control method, which integrates the residual sensory and motor abilities of the nonparalyzed upper body with the artificially controlled paralyzed lower extremities. Experiments, in which one intact and one paraplegic subject balanced under various experimental conditions while being constrained in a mechanical device named MRF, have shown the following:

- ankle stiffness of 8 Nm/° was adequate for comfort standing;
- both subjects were able to recover from disturbances acting in the artificial ankle joint;
- cognitive feedback was not prerequisite but it significantly enhanced the balancing abilities of both subjects

The experimental results also suggests that the MRF could be used for therapy and balance training of paraplegics.

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Zlatko Matjačić, for a photograph and biography, see this issue, p. 138.

Tadej Bajd (SM'91), for a photograph and biography, see this issue, p. 138.