Static and Dynamic Evaluation of the Influence of Supplementary Hip-Joint Stiffness on Crutch-Supported Paraplegic Stance

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Abstract—Paraplegic persons can stand with hip-knee-ankle-foot orthoses (HKAFO) and crutches. However, current HKAFOs restrict body movement extensively, which may impede functional upper-body movements. A more compliant body support using a more compliant orthosis or well-controlled functional electrical stimulation system may increase freedom of movement to the user, but should not impede stability and required arm support. In the current study, we investigated the consequences of varying stiffness applied at the hip to postural stability and required crutch force during paraplegic stance. Experiments were performed on five paraplegic persons with spinal cord lesions varying from T1 to T12. Static postures and dynamic responses to perturbations were tested for varying hip stiffness and crutch placements. The minimal hip-joint stiffness for stable stance appeared to depend on lesion level. In contrast to the predictions of a previous modeling study, no statistically significant influences of hip-joint stiffness or crutch-to-foot distance on posture and applied crutch forces were found. It is hypothesized that the main reasons of this discrepancy are the active upper-body efforts the paraplegic HKAFO users are still able to exert and the remaining flexibility of the upper trunk and shoulder region, which is present despite the restrictions of the orthosis.

Index Terms—Biomechanics, control systems, data processing, orthosis, paraplegic standing.

I. INTRODUCTION

MANY researchers have emphasized the importance of controlling paraplegic stance. Paraplegic individuals enjoy significant physiological [1] as well as physical benefits [2] from standing.

Many studies [3], [4] show that the often-employed techniques of functional electrical stimulation (FES) are not sufficiently effective to supply the large stabilizing moments that are required around the hip and ankle joints during paraplegic stance. Straightforward mechanical control strategy of

paraplegic stance is, as yet, essential. Matjacic reported on the application of additional ankle-joint stiffness combined with voluntary trunk control as a tool to realize arm-free paraplegic standing in a standing frame [5], [6]. A condition for this approach was sufficient voluntary control over the trunk, so only subjects with low thoracic lesions would benefit. An experiment on one (T12) paraplegic subject showed that $8 \text{ N} \cdot \text{m}^\circ$ of additional ankle-joint stiffness was necessary to obtain arm-free paraplegic standing.

Practical application of FES in balancing unsupported paraplegic subjects is despite its promises difficult for several reasons. In the first place, the many degrees of freedom in the human body are hard to control since it is difficult to stimulate a sufficient number of muscles selectively. Secondly, using electrical stimulation, the moments that can be generated around joints are limited [7]. Moreover, since balance control using FES implies stiffening the ankle and hip joints by co-contraction of antagonist muscle pairs, such strategies quickly lead to occurrence of muscle fatigue. Literature shows that standing performance for paraplegic subjects improves significantly when using hands and balancing aids [8]. Therefore, other studies have focused on balance control using an orthosis and supporting devices like crutches although sometimes in combination with FES [9].

However, purely mechanical control has its limits and disadvantages. Shoulder and wrist problems due to excessive and prolonged crutch loading are important factors that limit standing performance of paraplegic subjects [10], [11]. Therefore, in the design of control systems for crutch-supported paraplegic stance, the applied crutch forces should be minimized. The feasibility of intelligent FES control systems minimizing applied arm forces was reported in literature [12], [13]. A drawback of these approaches is that additional sensory information is needed as feedback signal. In practical applications, however, this additional sensory information may be difficult to measure and process.

In the current study, we aimed at stabilizing crutch-supported stance for paraplegic subjects for a wide range of lesion levels. Furthermore, our goal was to investigate the prerequisites for balancing paraplegics in practical situations: The subjects should be able to stand stable for a prolonged period of time, yet with enough freedom of movement to perform daily life tasks like handling objects and interacting with the environment. Paraplegics are able to stand stable in a knee-ankle-foot (KAFO) orthosis [14], although their freedom of movement is

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very restricted. Can stable crutch-supported stance be realized with less restriction of movement?

In a previous study [15], we modeled the effect of controlling crutch-supported paraplegic stance by adding mechanical hip-joint stiffness rather than ankle-joint stiffness [5], [6]. Less additional stiffness is then required around the hip compared to the ankle. Furthermore, adding ankle-joint stiffness is limited due to the danger of heel or toe lifting. Simulations predicted substantial differences in static and dynamic behavior when varying the amount of additional hip-joint stiffness, the hip joint offset angle or the placement of the crutches [15]. Both decreasing the crutch-to-foot distance and decreasing the supplementary hip-joint stiffness were predicted to obtain more erect static postures. Furthermore, these simulation results predicted that smaller crutch-to-foot distances would yield less arm forces. Also, for decreasing supplementary hip-joint stiffness, the arm forces were predicted to become less.

The goal of the current experimental study was to answer the following research questions.

- 1) What is the minimal required hip-joint stiffness to yield stable crutch-supported paraplegic standing?
- 2) What is the effect of changing the additional hip-joint stiffness on balance control under static and dynamic circumstances?
- 3) What is the effect of crutch placement on balance control under static and dynamic circumstances?

Experiments were performed with five paraplegic subjects, with lesion heights varying from T1 to T12. The additional hip-joint stiffness was realized by attaching passive springs to the orthosis worn by the subjects. Static standing tests and dynamic balance disturbance trials could thus be performed.

The experimental results will be compared with the model predictions and interindividual differences in the results will be analyzed. The answers to the research questions are expected to yield important knowledge relevant for the design of systems to enable paraplegics to stand in a functional way.

II. METHODS

A. Subjects

Measurements were carried out in five paraplegic subjects. Weights, lesion heights, and sex of the subjects are listed in Table I. All subjects had a complete thoracic spinal-cord lesion. The subject group was a convenience sample, which was heterogeneous with respect to lesion level. One subject had a low-thoracic lesion (T12), one subject a high-thoracic lesion (T1) and three subjects a lesion at intermediate height (T5–T9). All subjects signed an informed consent and the experiments were performed according to a protocol, which was approved by the local medical ethical committee.

B. Experimental Setup

Fig. 1 shows a schematic drawing of the experimental setup. Fig. 2 shows a photograph of one of the subjects in the experimental setup.

The subject was placed on a force plate (OR6-5 series, Advanced Mechanical Technology, Inc., Newton, MA), to measure the ground reaction forces and the center of pressure

TABLE I
CHARACTERISTICS OF THE SUBJECTS PARTICIPATING IN THE EXPERIMENTS.
THE SUBJECTS ARE NUMBERED IN ORDER OF LESION HEIGHT, FROM HIGH (T1) TO LOW (T12)

Subject #	Lesion height	Weight [N]	sex
1	T1	900	M
2	T5/T6	750	F
3	T8	677	F
4	T9	1025	M
5	T12	820	M

Fig. 1. Schematic representation of the experimental setup. The subject is placed on a force plate (1), while his/her left crutch (2) is supported by another force plate (1). The other crutch is placed parallel to the first crutch on the ground. The subject is wearing an orthosis (3) equipped with a passive spring over the hip joint (4) to provide for hip-joint stiffness. The knee joints are passive locked while the ankles are free. A safety belt (5) prevents the subject from falling. Force perturbations (push and pull forces) are given by a force applicator (6) that is driven by a pneumatic cylinder (7). Vicon markers are attached to the subject to monitor his/her movements in response to disturbances.

(CoP). One of the subject's crutches was placed on another force plate, to measure the ground reaction forces under the crutches during stable stance and during the balance perturbation trials. The subjects were all secured by means of ropes to a safety frame, to prevent the risk of falling. The placement of the force plates, the feet, and the crutches are shown in Fig. 2. The crutches were instrumented with miniaturized load cells (LM-100KA, Kyowa Electronic Instruments Ltd., Tokyo, Japan) to measure the axial crutch forces applied. During the experiments, the subject was wearing a modified ARGO steeper (Fig. 2) hip-knee-ankle-foot orthosis (HKAFO), including a trunk brace with a high strap over the chest. The following several modifications were made to the orthosis:

- 1) reciprocal hip linkage cable was removed to enable free movement of the upper body;

Fig. 2. Photograph of one of the subjects standing in the experimental setup, preparing for a perturbation trial.

TABLE II
MECHANICAL SPRINGS THAT WERE USED IN THE EXPERIMENTS AND THEIR STIFFNESS VALUES IN $\text{N} \cdot \text{m/rad}$

Spring	Stiffness [Nm/rad]
K_0	0
K_1	41
K_2	68
K_3	126
K_4	208
K_5	313

- 2) exchangeable mechanical springs were placed over the hip joints on both sides, to provide for the variable hip-joint stiffness;
- 3) knee joints were mechanically locked;
- 4) hinge to enable free-ankle motion in the sagittal plane replaced the ankle joint of the orthosis;
- 5) lengths of the metal bars connecting the joints were made adjustable to account for different body dimensions of the subjects.

The application of springs with different wire diameters realized different hip-joint stiffnesses. Table II specifies the stiffnesses of the manufactured springs, as they were measured on the orthosis. The stiffness values were categorized from $K_0 = 0$ to $K_5 = 313 \text{ N} \cdot \text{m/rad}$. The hip-joint offset angle (the hip-joint angle at which the springs did not generate a moment around the hip) could be varied from $\pm 0.35 \text{ rad}$ to $\pm 0.35 \text{ rad}$. A hip-joint offset angle of 0 rad means that the leg segments were aligned with the trunk segment for the unloaded orthosis. The attachment of the springs to the orthosis was such that the spring could be replaced quickly, without removing the orthosis from the subject.

A force-applicator device was used to generate disturbance forces during stance. For this purpose, a pneumatic cylinder was used [16] to generate push and pull forces that could be changed by changing the air pressure. This perturbation device only made mechanical contact to the orthosis of the subject during a perturbation, so as to ensure that the postural dynamics of the subject were not influenced before and after the perturbation.

During the experiments, both quiet standing trials and balance perturbation trials were performed. The applied force was measured with a load cell. Movements were measured and assessed using five Vicon cameras (Oxford Metrics Ltd., Oxford, U.K.). Reflective markers were attached to the orthosis, to measure body movements, to one of the crutches, to measure crutch displacement and to the force applicator, to measure the angle under which the disturbance forces were applied. The markers that were placed on the orthosis were placed on both sides: on the ankles, hip joints, and shoulders of the subject. All Video data was sampled at a 50-Hz sampling frequency, whereas all analog data was sampled at 200 Hz.

C. Experimental Protocol

The load cells in the crutches and on the perturbation device were calibrated with zero and with 20 kg weights. In a separate measurement, it was ascertained that the load cell signals varied linearly with the applied weight. The Vicon system was calibrated using standard static and dynamic calibration trials. Unloaded force plate measurements were performed to determine the offsets of the various force plate channels.

After fitting the orthosis to the subject and aligning the hip joint with the rotation axes of the hip joints of the orthosis, some quiet standing trials were performed to familiarize the subject with the setup and with crutch-supported standing. Initially, no spring was applied at the hip joint to determine the capability of the subjects to stand freely without supplementary hip-joint stiffness. Subsequently, the stiffness was increased stepwise, to find the minimal stiffness required to obtain stable stance. The offset angle that was used in the measurements was determined to be that offset angle that allowed the subjects to stand most comfortably.

After the familiarization period, the disturbance trials were carried out. The subjects were instructed to stand comfortably placing the tip of the crutches at a predetermined distance (R_1) from the feet. Two such distances were used, referred to as “close” (approximately 30 cm) and “far” (approximately 50 cm). In this stage, subjects were instructed to minimize the applied crutch forces by standing as erect, yet as comfortably, as possible.

For every subject, disturbance trials were performed with several hip-joint stiffness values K_S . For each K_S , disturbance trials were performed with two different crutch-to-foot distances R_1 . For each of these distances, one trial was performed using push forces and one using pull forces. This yields a total of four trials per hip-joint stiffness value. During a single disturbance trial, six disturbances were applied at three force levels and two different durations of application. Every disturbance trial in which six push forces were applied was immediately followed

Fig. 3. Measured disturbance forces N and hip-joint angle responses (degrees) for push (A, C) and pull (B, D) trials, plotted against time. The quantities A in (a) and (b) represent the impulse of the applied force pulses. The gray boxes in (c) and (d) define the 1-s time intervals over which the average hip-joint angles μ were calculated to specify the postures in between perturbations. The quantities Δ denote the hip-joint angle response amplitudes for each particular disturbance applied.

by a trial in which six pull forces were applied, while keeping the crutch placement and applied hip-joint stiffness unchanged. Before and after each disturbance, some seconds of unperturbed standing were recorded. Before the measurements, the subjects were instructed to try to return to the original posture that they had before a particular disturbance force was applied.

D. Data Processing

All signals measured were filtered using a third-order 10-Hz lowpass filter, to eliminate high-frequency noise. From spectral analyses, it was made clear that no significant frequency components above 5 Hz were present in the power spectra of the recorded system. Prior to lowpass filtering, the analog signals (forces) that were originally sampled at 200 Hz were resampled to yield signals with 50-Hz sampling frequency, in accordance with the sampling frequency of the Vicon system.

From the measured positions of the Vicon markers, all angles of interest could be calculated. The following angular information was obtained from the Vicon data:

- 1) ankle-joint angles left and right ($\varphi_{\text{anklel}}/\varphi_{\text{ankler}}$), calculated from markers on the ankles and hip joints;
- 2) hip-joint angles left and right ($\varphi_{\text{hipl}}/\varphi_{\text{hipr}}$), calculated from markers on the ankles, the hip joints and the shoulder joints;
- 3) crutch angle (only for the left crutch);
- 4) angle under which the disturbance force was applied;
- 5) hip-joint torsion angle, measured from the markers at the left and right hip joint.

Furthermore, the crutch length (d_3) and the crutch-to-foot distance R_1 were calculated from Vicon markers positions. The hip and shoulder moments generated by the crutch forces were calculated from the ground reaction force under the crutch, its point of application, and the shortest distance from the ground reaction force vector to hip joint and shoulder, respectively.

From preliminary experiments, it was clear that some markers were less likely to be fully monitored over the entire duration of the trials. This was especially the case for the marker at the right hip joint. The safety frame that was placed over the subject sometimes obscured this marker. To deal with this, the position of that particular marker was reconstructed from other marker positions: From the time intervals in which the right hip joint marker was properly monitored, the position of the right ankle was estimated and assumed static over the entire trial. Given this, the right hip-joint marker could be reconstructed from the estimated right ankle joint and the right lower-leg marker. For the time intervals in which the right hip-joint marker was not available, the coordinates of the reconstructed hip-joint marker in the same time intervals were taken instead.

On the basis of reconstruction of hip-joint marker positions in trial sections where the right hip-joint marker did not disappear, the root mean square (rms) error of hip-joint marker position was estimated to be approximately 3 mm in the y and z directions and 6 mm in the x direction. The resulting estimation of hip-joint angle had an rms error of 0.02 rad. From this, it can be concluded that the marker reconstruction algorithm is a feasible way to reconstruct missing markers.

E. Assessment of Static and Dynamic Behavior

The hip-joint angle was taken as a measure of the static postures. Since Vicon markers were placed on both sides of the subject's body, the hip-joint angle could be calculated from the markers placed on either side of the body, yielding the hip-joint angles at the left and right sides φ_{hipl} and φ_{hipr} . The hip-joint angle φ_{hip} was defined as the average of φ_{hipl} and φ_{hipr} .

The procedure that was followed in the quantification of static and dynamic behavior is explained by Fig. 3, that represents two different hip-joint angle responses during a disturbance trial measured from one of the paraplegic subjects. Fig. 3(c) illus-

trates a trial in which push disturbances were applied. Fig. 3(d) illustrates a trial with pull disturbances. Fig. 3(a) and (3b) show the force/time characteristics of the disturbance force applied, for push and pull disturbances, respectively.

As can be seen from Fig. 3, several seconds of unperturbed stance were recorded between the individual disturbances that were applied. Fig. 3 shows that for both push and pull disturbance trials the measured responses prior to and after each perturbation are not equal. This phenomenon was consistent over all response signals that were measured during the experiments.

Before the onset of each individual perturbation force, a 1-s time window was defined over which to average the measured response. The gray boxes in Fig. 3(c) and (d) illustrate these time windows. For a total of six perturbation forces applied per trial, this yielded six averaged values μ_1 – μ_6 . We characterized static behavior by the mean and standard deviation of the values μ_1 – μ_6 . For the hip-joint angle responses plotted in Fig. 3, this procedure yielded two measures of the mean static hip-joint postures, one for the push perturbation trial and one for the pull perturbation trial.

The same procedure was applied to calculate mean and standard deviation of the static crutch force $F_{\text{crutch},s}$, the static hip-joint moment $M_{\text{hip},s}$, and the static shoulder joint moment $M_{\text{shoulder},s}$ exerted by the crutch forces.

Fig. 3 also illustrates the method to study dynamic behavior: the parameters A_i represent the force impulse (time integral of force) delivered by the i th perturbation. The parameters Δ_i represent amplitude of the response due to application of the i th perturbation force, and are defined as the difference between the response peak value after the perturbation and the value at the time of onset of the perturbation. When studying the dynamics, parameters A_i are considered the input variables and Δ_i the output variables. Linear regression with zero offset was used to calculate the gain of the best linear fit and the correlation coefficient between the input and output variables. Mean and standard deviations of the estimated gains were calculated for the hip-joint angle (G_A), crutch force (G_f), hip-joint moment (G_{hm}), and shoulder joint moment (G_{sm}).

F. Statistical Tests

All static and dynamic parameters were compared within each subject and over all subjects. Two-way ANOVA statistical tests for repeated measures were performed to evaluate the dependencies on the factors hip-joint stiffness K_S and crutch-to-foot distance R_1 . For all statistical tests (static and dynamic) reported in this article, the stiffness values were categorized, according to Table II, into six levels of stiffness $K_0 \dots K_5$. The crutch-to-foot distance R_1 was categorized into “close” and “far,” as mentioned before. All statistical tests were performed using SigmaStat Software. Differences were considered significant for $p < 0.05$.

III. RESULTS

A. Minimal Stiffness as a Function of Lesion Height

For every subject, it was determined what minimal stiffness K_S was required to obtain stable stance. This was done by stepwise applying more stiffness (starting without stiffness) and

TABLE III
MINIMAL LEVEL OF STIFFNESS NECESSARY TO ENABLE EACH INDIVIDUAL SUBJECT TO STAND

Subject #	Lesion height	Min. level of stiffness to stand stably
1	T1	K_3
2	T5/T6	K_2
3	T8	K_2
4	T9	K_2
5	T12	K_0

subsequently checking whether the subject was able to stand stable. The results are presented in Table III.

Of these five subjects, the one with the highest lesion (T1) needed the highest additional hip-joint stiffness to stand stable. This may be understood considering a higher lesion implies less control over the trunk, requiring a larger effort to stabilize stance. The other four subjects do not contradict this, although the number of subjects is too small to reach any statistically significant results.

B. Static Results

Fig. 4(a) gives the static crutch force $F_{\text{crutch},s}$, with the standard deviations, for all subjects, for all stiffness values used and both crutch-to-foot distances R_1 . In order compare the various subjects the crutch force was normalized on the individual subject's body weight.

It can be seen from Fig. 4(a) that the results for the different subjects and the different conditions are not consistent. While for one subject the crutch force decreases when placing the crutches further away (subject 5) the result is opposite for other subjects (subjects 1 and 4). For some subjects, no clear effect is noticeable. Furthermore, no consistent effect of changing stiffness is observed.

Fig. 4(b) gives the static hip-joint angles $\varphi_{\text{hip},s}$ and standard deviations belonging to the static crutch forces that were plotted in Fig. 4(a). Positive hip-joint angles represent hip flexion, whereas negative hip-joint angles represent hip extension. From Fig. 4(b), it can be seen that the effect of stiffness and crutch-to-foot distance on the static postures also is not consistent. For some subjects, the postures become more erect than for others when applying more stiffness. The effect of crutch placement on the equilibrium postures also differs from subject to subject.

Furthermore, it can be seen that, in some cases, the standard deviations of the various hip-joint angles and crutch forces can be big. Apparently, the paraplegic subjects are still able to influence their preferred posture voluntarily by using more or less arm force, or possibly shoulder moments.

Fig. 4(c) shows the influence of hip-joint stiffness K_S and crutch-to-foot distance R_1 on the static hip-joint moment $M_{\text{hip},s}$ induced by the crutch force. As is the case for the crutch force, the hip-joint moment has been normalized on the subject's weight, to make comparison between subjects possible. There is a clear effect of changing the crutch-to-foot

Fig. 4. Averaged static data for all subjects and all disturbance trials. Plotted are the (a) mean crutch force as a percentage of body weight, (b) the mean hip-joint angle, and [(c) and (d)] the mean hip-joint and shoulder joint moments due to crutch forces. On the horizontal axes the stiffnesses applied on each subject are plotted, grouped by subject. The vertical lines separate data of different subjects. Light gray bars represent results of trials in which the crutches were placed close to the subjects, whereas dark gray bars represents results of trial with the crutches placed more distant. All bars represent averages and standard deviations.

distance on the contribution of the crutch force to the hip-joint moment. No clear difference is observed in results between the various subjects.

Fig. 4(d) shows the static shoulder joint moments $M_{shoulder,s}$ induced by the crutch force. It can be seen from Fig. 4(d), that the shoulder joint moment induced by the crutch force is not

negligible compared to the hip-joint moment induced by the crutch force. Furthermore, the shoulder joint moment is very variable. Possibly, the subjects were actively influencing their posture by using shoulder moments. This would imply that the subject having most voluntary control over the shoulder joint (subject with lowest lesion), generates the largest shoulder mo-

Fig. 5. Input/output characteristics of four disturbance trials. The horizontal axis plots represents the disturbance force impulse A Ns/kg, whereas the vertical axis represents the crutch force amplitude Δ N/kg, both divided by the subject's mass. The legend shows the symbols used in the graph for every individual trial. In addition, the linear approximations obtained from linear regression analysis are plotted. The gains G and correlation coefficients R for the different regression lines are: crutches close— $K = K2$: $G = 1.16$, $R = 0.95$; crutches far— $K = K2$: $G = 1.06$, $R = 0.97$; crutches close— $K = K5$: $G = 1.60$, $R = 0.95$; crutches far— $K = K5$: $G = 1.36$, $R = 0.99$.

ments. From Fig. 4(d), it can be seen that the fifth subject having the lowest lesion indeed uses the largest shoulder moment.

C. Significance of Static Measurement Results

Statistical testing of the static results showed that crutch-to-foot distance had a significant effect on the static hip-joint moment contribution of crutch force. No other significant effect of hip-joint stiffness or crutch-to-foot distance was found on the evaluated quantities. No significant interaction effects between influence of hip-joint stiffness and crutch-to-foot distance was found either.

D. Result of Dynamic Measurements

As mentioned in the previous section, the dynamic behavior for different stiffnesses K_S or crutch-to-foot distance R_1 was characterized by the mean and standard deviations of the gains of the linearized input/output characteristics obtained from linear regression. Input is the disturbance force impulse and output is the (angle, force, hip, or shoulder moment) response amplitude. Fig. 5 gives an impression of a typical input/output characteristic measured during one of the experiments. In Fig. 5, the crutch-force response is plotted against the disturbance force impulse. Furthermore, the linear regression lines are plotted. The regression results are given in the caption of Fig. 5.

All correlation coefficients in this example are larger than 0.95. Other measurements for different subjects and different output show similar results. Therefore, characterizing the dynamic analysis results using linear regression of input/output

characteristics was justified. Gains were calculated for the crutch force (G_f), hip-joint angle (G_A), hip-joint moment (G_{hm}), and shoulder moment (G_{sm}). Fig. 6 gives the results for both crutch-to-foot distances and all hip-joint stiffnesses K_S . The standard deviations of the gains estimated from the linear regression analysis are also plotted in Fig. 6.

No systematic influence of additional hip-joint stiffness on the crutch force gain G_f can be seen in Fig. 6(a). Also, the change in G_f for different subjects seems negligible. However, consistent decrease of G_f when placing the crutches further away can be clearly seen. In the case of hip-joint angle [Fig. 6(b)] the effect of crutch placement on G_A is not consistent, neither is the effect of changing the stiffness K_S on G_A . The estimated gains for hip-joint moment and shoulder joint moment induced by the crutch forces, G_{hm} and G_{sm} , are plotted in Fig. 6(c) and (d), respectively. These figures suggest the gains G_{sm} are not negligible in comparison to G_{hm} .

E. Significance of Dynamic Measurement Results

Statistical testing of the dynamic results showed that crutch-to-foot distance had a significant effect on the crutch force gain G_f . This means that in case of balance perturbation less additional crutch force is used in balance restoration when the crutches are placed further away. No other significant effect of hip-joint stiffness or crutch-to-foot distance was found on the evaluated quantities. No significant interaction effects between influence of hip-joint stiffness and crutch-to-foot distance was found either.

Fig. 6. Averaged dynamic data for all subjects and all disturbance trials. Plotted are the (a) mean normalized crutch force gain 1/Ns as a percentage of body weight, (b) the mean hip-joint angle gain rad/Ns, (c) the mean hip-joint moment gain m/kgs, and (d) the mean shoulder-joint moment gain m/kgs due to crutch forces. On the horizontal axes, the stiffnesses applied on each subject are plotted, grouped by subject. The vertical lines separate data of different subjects. Light gray bars represent results of trials in which the crutches were placed close to the subjects, whereas dark gray bars represents results of trial with the crutches placed more distant. All bars represent averages and standard deviations.

IV. DISCUSSION AND CONCLUSION

In a previous study [15], a biomechanical model of crutch-supported paraplegic stance was made. Using biomechanical analyses, it was hypothesized there that by stabilizing the hip joint using additional hip-joint stiffness, crutch-supported paraplegic stance could be achieved. The effects of adding more stiffness and changing the crutch positions were predicted using

the model: considerable changes in static equilibrium postures and balance stabilization after perturbations were predicted. In the current study, these model predictions were experimentally tested.

From the results presented in this paper, we concluded that using additional hip-joint stiffness it is possible to realize stable crutch-supported paraplegic stance. All could stand stable and all could restore balance after perturbations. The measurement

results are not contrary to the intuitive motion that subjects with a higher thoracic lesion require more additional hip-joint stiffness to stand stable. The subject with the lowest lesion (T12) was able to stand freely, even without crutch support and additional stiffness, whereas the subject with the highest lesion needed considerably larger hip-joint stiffness. Since the number of paraplegic subjects that participated in the measurements was limited, no statistical analysis was possible of the influence of lesion height on minimal hip-joint stiffness needed to control stance. However, the model did not predict such a dependency on lesion level, possibly because no voluntary control was assumed of the hip joint and a stiff pelvis and trunk. The beneficial effects of adding hip-joint stiffness were acknowledged by all subjects: even though adding more hip-joint stiffness did not result in significant different dynamic responses, subjects felt more secure when the hip joints were made stiffer.

A significant effect of the crutch-to-foot distance on the hip-joint moment generated by the crutches was found, although the influence on the crutch force F_{crutch} and hip-joint moment angle φ_{hip} was not significant. During balance recovery after perturbations, significantly less stabilizing crutch force is used to restore balance when the crutches are placed further away.

However, no significant effect of changing the hip-joint stiffness K_S on static posture, crutch forces, and hip and shoulder moments was found. Also, the ability to cope with disturbance forces does not change with stiffness. This is in contrast with our simulation study [15] in which hip-joint stiffness was predicted to influence the postures and crutch forces under static and dynamic conditions. Furthermore, the static postures that were observed were mainly postures in which the hip joint was in flexion. When using an insufficient amount of hip-joint stiffness the subjects were not able to stand stable. In those situations, the flexion or extension stop in the orthosis prevented the subject from becoming unstable. This is in agreement with measurements performed by Baardman *et al.* [16], where paraplegic subjects wearing different orthosis showed different standing postures, depending on whether the flexion stop or extension stop of the orthosis was used in balancing.

In summary, it can be said that even for a heterogeneous group of subjects with varying lesion heights, weights, and other physical properties, hardly any statistically significant influence of crutch position and hip-joint stiffness was observed. Even though the subject with the lowest lesion was able to stand stable, unsupported and without additional stiffness around the hip, he used the same level of upper body effort as the other subjects in the trials where crutches were used. In addition, the results of the dynamic measurements do not show any significant difference in applied crutch forces between the individual subjects.

It is thought that there may be two main reasons for the lack of significant influence of crutch position and hip-joint stiffness on static and dynamic behavior: voluntarily applied arm forces and nonrigidity of the upper body, contrary to model assumptions [15]. These two effects are closely related: When the upper body, shoulders, and arms are completely rigid, no voluntarily arm forces can be applied. In this situation, only a shoulder moment could be applied.

During the experiments, the subjects could control balance to a certain extent by using forces exerted on the crutches and moments generated around the shoulder. In this way, the subjects seem to actively influence their postures and kinetics. In contrast, the model assumed that a subject with an orthosis behaves like a passive system. The exerted crutch forces calculated from the model were merely a result of the postures that were found. In the measurements, however, a close interaction between posture and applied crutch force was not found. The voluntary activity (applied crutch force) under static and dynamic conditions was not influenced by posture or crutch-to-foot distance.

The role of upper body effort in balancing was also studied by Matjacic *et al.* [6]. They showed that unsupported paraplegic stance could be achieved by a combination of additional stiffness around the ankle joint and voluntary stabilizing movements of the upper body. Voluntary control of the upper body was a condition for this approach. In our experiments, the T12 subject was able to stand freely as well, even without additional ankle-joint stiffness. In this case, only voluntary upper body effort was used to stabilize. Other studies have shown the important role of the upper body in control of balance—flexibility of movement of the spine and coordinated effort of lower back muscles are important in postural stability and spinal loading [17], [18].

Another aspect that could explain the results from this study is the fact that, contrary to what was modeled, the upper body was not rigid. Even though the subjects were all tightly secured and strapped to the orthosis, trunk and pelvis were not fully immobilized. This does not mean that the trunk orthosis part is not required. It is expected to contribute in enabling effective voluntary influencing of posture. Keeping this in mind, it can be expected that the subject with the lowest lesion used larger shoulder moments compared to the rest of the subjects. The results indicate that this is true. Part of this shoulder moment was probably caused by the fact that a large part of the upper body was under voluntary control.

The overall conclusion from the presented results is that additional hip-joint stiffness results in stable crutch-supported paraplegic standing. Although not statistically tested, measurements indicate that subjects with higher thoracic lesions needed more additional hip-joint stiffness to be able to stand stable. The absence of statistically significant influences of stiffness or crutch placement on crutch forces and balance, contrary to what simulations predicted, indicates that some important aspects of crutch support paraplegic standing were not modeled properly. It is thought that the effect of trunk nonrigidity and the inherent trunk movement, forces and moments, were underestimated in the model. Further research is required to understand the influence of trunk flexibility and voluntary trunk control on stabilizing crutch-supported paraplegic stance.

Certain orthotic components may be substituted by electrical stimulation. For example, locking of the knee joint and its timing can be achieved by using a simple control system needing only sensory information about the current state of knee locking [19]. Increasing stabilization in the trunk may also be done using electrical stimulation. Since electrical stimulation of the trunk muscles will have a stiffening effect on the trunk, applying electrical stimulation of trunk muscles could well decrease the additional hip-joint stiffness required for stable crutch-supported stance.

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