A Model-Based Approach to Stabilizing Crutch Supported Paraplegic Standing by Artificial Hip Joint Stiffness

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***Abstract*—The prerequisites for stable crutch supported standing were analyzed in this paper. For this purpose, a biomechanical model of crutch supported paraplegic stance was developed assuming the patient was standing with extended knees. When using crutches during stance, the crutches will put a position constraint on the shoulder, thus reducing the number of degrees of freedom. Additional hip-joint stiffness was applied to stabilize the hip joint and, therefore, to stabilize stance. The required hip-joint stiffness for changing crutch placement and hip-joint offset angle was studied under static and dynamic conditions. Modeling results indicate that, by using additional hip-joint stiffness, stable crutch supported paraplegic standing can be achieved, both under static as well as dynamic situations. The static equilibrium postures and the stability under perturbations were calculated to be dependent on crutch placement and stiffness applied. However, postures in which the hip joint was in extension (C postures) appeared to the most stable postures. Applying at least 60 N · m/rad hip-joint stiffness gave stable equilibrium postures in all cases. Choosing appropriate hip-joint offset angles, the static equilibrium postures changed to more erect postures, without causing instability or excessive arm forces to occur.**

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

I. Introduction

RECENTLY, there has been an increasing interest in the ability to control paraplegic stance. If individuals with paraplegia were able to stand comfortably for a prolonged period of time, it would have many therapeutic, psychological, and practical advantages [1], [2].

Much of the research regarding the control of paraplegic standing has been focused on the use of artificial stimulation to activate paralyzed muscles and to partially restore motor function [1]. Activation of muscles either around the ankle (ankle strategy), or the hip (hip strategy), or both assists the healthy individual in keeping the body's center of gravity within its base of support [3] when stability is perturbed [4]. In a study by Kuo [5], the role of combined hip and ankle strategy during postural control was implemented in a model as an optimal feedback postural control system.

In several studies, successful standing trials with paraplegic subjects were obtained both by applying electrical stimulation of quadriceps muscles and by using support devices like standing frames, orthoses with reciprocal linkage, and crutches [6]–[8]. To minimize muscle fatigue, closed-loop control of stimulation [9], [10] or posture switching [11] was used. For paraplegics standing with an orthosis, the use balancing aids greatly improved standing performance, especially in the saggital plane [12]. This may be because of the limited strength of stimulated paralyzed muscles, resulting in limited possibilities to obtain stable postures using electrical stimulation only [13].

Literature also shows evidence that the role of arm support is a major factor in controlling paraplegic stance. Excessive arm forces may make paraplegic standing exhaustive, while stiffening joints too much may limit freedom of movement. Recently, a number of studies were performed that tried to minimize the applied arm forces during different tasks [14], [15]. Both studies were model based and predict improved movements and reduced arm forces in functional electrical stimulation (FES)-supported standing up and standing. However, experiments to support this still have to be performed.

An alternative way to stabilize posture was proposed by Matjacic *et al.* [16]. In a modeling study, increased ankle-joint stiffness in combination with voluntary trunk effort was analyzed to obtain stable paraplegic stance. The conclusion was that individuals with paraplegia would be able to stand freely, only balancing by voluntary trunk movements, when a certain amount of ankle-joint stiffness was applied. Experiments with one paraplegic subject [17] showed the feasibility of the proposed control strategy. This experimental study, however, focused on arm-free paraplegic standing using a standing frame, which is contrary to our approach where we aim at stabilizing crutch supported standing.

Even though current literature provides many different proposed control strategies for paraplegic standing, there is a limited understanding of the biomechanical aspects of paraplegic standing. For instance, if a paraplegic person is standing supported by crutches, what is the prerequisite to balance the joints? Furthermore, if balance is perturbed, what extra control mechanism has to be incorporated in order to cope with these perturbations? How can this control mechanism be implemented in order

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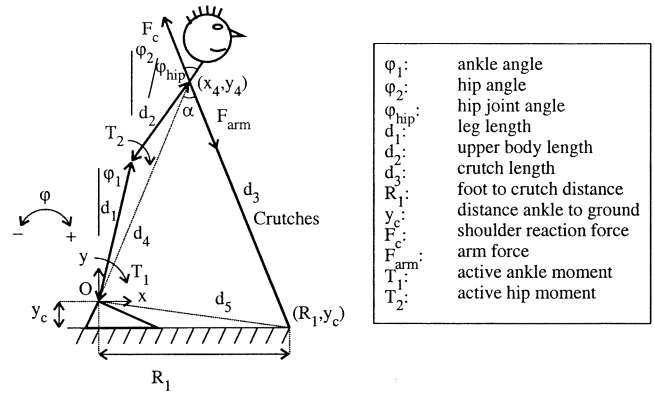


Fig. 1. Double inverted pendulum model of crutch-assisted paraplegic standing. The knee is assumed to be locked. In the lower link, the shanks and thighs of both legs are lumped together. The length of the lower link is d_1. The upper link consists of the upper body, arms, and head lumped together and has length d_2. The crutches have length d_3, are assumed massless, and only force the shoulder position to be constrained. The tip of the crutches is at a horizontal distance R_1 of the ankle joint. The distances d_4, d_5, and angle α are only used in calculations and have no functional meaning. F_c represents the constraint force, and F_{arm} represents the arm force. These two forces are equal in magnitude and opposite in direction.

to minimize the applied arm force? It is our aim to answer these questions in this modeling study by focusing on the effects of additional joint stiffness, where paraplegic stance should meet the following criteria.

1) Standing is stable, making the application of additional joint stiffness necessary since previous experiments showed that crutch support alone is not sufficient [19].
2) The arm force exerted on the crutches is minimal, i.e., arm forces are only used to assist balance. This is necessary to avoid fatigue in arms and hands and to avoid overstress of the shoulder [18].
3) The control system gives the user maximum freedom of movement in order not to limit activities while standing. Therefore, additional joint stiffness should not be too high.

II. Analysis

A. Model of Crutch Supported Paraplegic Standing

For analysis of the system, a two-link segmental model was constructed. The first link represents the shanks and thighs of both legs lumped together. The second link consists of the head, arms, and trunk (HAT). The model was two-dimensional, considering motion in the sagittal plane only. The feet were included in the model as well, but since they are considered to be static, they do not influence the system's dynamics but simply define the position of the ankle joint. Muscles were not included in this model. Externally applied moments to the ankle joint and hip joint were represented as ideal joint moments. The influences of passive damping and elasticity were neglected. Fig. 1 shows the model.

Instead of modeling the arms and crutches as a third link with its own, mass dependent, dynamics, they were considered massless, only imposing a position constraint on the shoulder, thus forming a closed kinematic chain. For equilibrium, the reaction force acting at the crutch/floor interface was assumed to be the same as the arm force exerted by the subject on the crutches.

In Fig. 1, φ_1 and φ_2 denote ankle and trunk angle, respectively. The hip-joint angle φ_{hip} is defined as $\varphi_2-\varphi_1$. Positive φ_1 indicates dorsal flexion and positive φ_{hip} indicates hip flexion. Inputs to the model are the externally applied ankle and hip moments T_1 and T_2. Both moments are defined positive when applied clockwise.

Equations of motion were derived using Lagrange's method [20]. The Lagrange equations were extended with the influence of the additional force F_c (see Fig. 1), which keeps the distance between (x_4 and y_4) and (R_1 and y_c) in Fig. 1 constant at a distance d_3, which is the crutch length

$$\begin{bmatrix} Q_1 \\ Q_2 \end{bmatrix} = \mathrm{M}(\varphi) \cdot \begin{bmatrix} \ddot{\varphi}_1 \\ \ddot{\varphi}_2 \end{bmatrix} + \mathrm{V}(\varphi_1, \varphi_2, \dot{\varphi}_1, \dot{\varphi}_2) + \mathrm{G}(\varphi_1, \varphi_2) - C(\varphi_1, \varphi_2) \cdot F_c. \tag{1}$$

$M, V,$ and G represent the inertia matrix, the coriolis matrix, and the gravitational matrix M. Vector $C(\varphi_1, \varphi_2)F_c$ represents the moment generated by the constraint force F_c acting at the shoulder. The position constraint equation can be formulated as follows:

$$(R_1 - d_1 \cdot \sin\varphi_1 - d_2 \cdot \sin\varphi_2)^2 + (y_c + d_1 \cdot \cos\varphi_1 + d_2 \cdot \cos\varphi_2)^2 = d_3^2. \tag{2}$$

Using a method previously formulated by Pandy and Berme [21] yet applied for our situation, the implicit solution of (2) for φ_1 and φ_2 was calculated. Introduction of the position constraint of the shoulder, causes the number of degrees of freedom of the system to be one.

B. Stability of Crutch Supported Paraplegic Standing

Stability of standing can be studied analyzing the potential energy of the system as a function of φ_1, φ_2, or φ_{hip}. Local minima of this function are stable equilibrium positions.

In an open-loop situation, where no control tools are applied, the potential energy function will show no local minima, and the system will, therefore, be unstable. Stabilization of the system requires stabilization of the only degree of freedom in the system by means of feedback control. This can be done by either stabilizing the ankle angle φ_1 or by stabilizing the hip-joint angle $\varphi_h (= \varphi_2 - \varphi_1)$ using additional joint stiffness. The potential energy $E_{p,\mathrm{stiff}}$ that is added to the system as a function of the joint angle $\varphi (= \varphi_{\mathrm{hip}}$ or $\varphi_1)$ equals

$$E_{p,\mathrm{stiff}} = \int_{\varphi_0}^{\varphi} K_S \varphi d\varphi = \frac{1}{2} \cdot K_S \cdot (\varphi - \varphi_0)^2. \tag{3}$$

The angle φ_0 represents the angle (hip or ankle) for which no moment is generated by the stiffness. Since $E_{p,\mathrm{stiff}}$ increases quadratically with the angle φ, it can be concluded that a large enough value for K_S will yield a total potential energy $E_{p,\mathrm{tot}}$ that indeed gives a local minimum for a certain angle. The system will be in equilibrium for that angle.

In dynamic simulations, not only joint stiffness but also joint damping K_D has to be added to prevent oscillations. Using a combination of joint stiffness and damping the following stabilizing moment is generated:

$$T_2 = K_S \cdot \{(\varphi_2 - \varphi_1) - \varphi_0\} + K_D \cdot (\dot{\varphi}_2 - \dot{\varphi}_1) \quad (4)$$

if additional hip-joint stiffness and damping is used. If additional ankle-joint stiffness is used, the stabilizing ankle moment becomes

$$T_1 = K_S \cdot (\varphi_1 - \varphi_0) + K_D \cdot \dot{\varphi}_1 \quad (5)$$

In (4) and (5), φ_0 represents the hip-joint angle or ankle-joint angle, respectively, for which the stiffness imposes no stabilizing moment. From this point, φ_0 will be referred to as the offset angle.

III. Simulation Procedures

The influence of various values of additional stiffness and damping on the stability of crutch supported standing was calculated in static and dynamic situations. Stability, posture, and exerted arm force were evaluated for varying stiffness K_S, damping K_D, and distance between the feet and crutches R_1.

In the static analyses, the equilibrium postures (defined by the local minimum occurring in the potential energy function), the static arm forces, and static joint moments were studied as a function of various changing system parameters: Stiffness K_S, Damping K_D, foot to crutch distance R_1, and offset angle (φ_0).

In the dynamic analyses, the disturbance handling capabilities of the equilibrium postures were investigated for various system parameter values. The stable equilibrium postures (i.e., values of φ_1, φ_2, and φ_{hip} in equilibrium) were perturbed using a pulse-shaped hip-joint moment $T_{2,\mathrm{dist}}$. The duration of the pulse was 1 s in all cases and was preceded by 1 s of no disturbance. The duration of the disturbance pulse was chosen to be in the same order of magnitude as the dynamic responses of the system to the perturbations. The resultant responses were examined using the same system parameters as in the static analysis. The damping K_D for a given stiffness K_S was determined by a small-signal analysis in the equilibrium posture. The model was linearized in equilibrium, resulting in a linear second-order system. By varying of the damping K_D for a fixed stiffness K_S the poles of the system were chosen such that a stable response with 4% overshoot (relative damping $= 0.7$) was to be expected. The following analyses were performed.

- *Determination of the maximal disturbance moment $T_{2,\mathrm{dist}}$ that still gives stable responses.* This was done for disturbance moments applied in both directions. A disturbance moment applied in clockwise direction (see Fig. 1) is referred to as a "flexion moment" because it forces the hip joint to flex. In analogy with this, counter clockwise disturbance moments are referred to as "extension moments."
- *Transient analysis of joint angles and reaction forces under various levels of perturbation.*

The model parameter values used in the simulations are listed in Table I. They are obtained from an actual person with paraplegia who participated in previous experiments.

TABLE I
MODEL PARAMETERS USED IN ALL SIMULATIONS THAT WERE PERFORMED. THESE PARAMETERS WERE BASED ON AN ACTUAL PARAPLEGIC SUBJECT'S MEASURES. THE MASSES m_f, m_1, AND m_2 WERE CALCULATED AS FRACTIONS OF THE TOTAL BODY MASS, USING ANTHROPOMETRICS [20]. THE LENGTHS d_{1a}, d_{1b} d_{2a}, d_{2b}, x_c, x_f, AND y_f WERE CALCULATED AS FRACTIONS OF THE SUBJECT'S LENGTH. THE MASS MOMENTS OF INTERTIA I_1 AND I_2 WERE CALCULATED, AGAIN BASED ON ANTHROPOMETRICS

	parameter	*value*
Foot parameters:	m_f	1.3 [kg]
	x_f	0.05 [m]
	y_f	0.05 [m]
	x_a	0.06 [m]
	x_b	0.18 [m]
	y_c	0.10 [m]
Lower body parameters:	m_l	20.9 [kg]
	d_{1a}	0.34 [m]
	d_{1b}	0.43 [m]
	I_1	1.29 [kgm^2]
Upper body parameters:	m_2	37.6 [kg]
	d_{2a}	0.26 [m]
	d_{2b}	0.21 [m]
	I_2	5.71 [kgm^2]
Other parameters:	R_1	0.3 [m]
	d_3	1.35 [m]

IV. Simulation Results

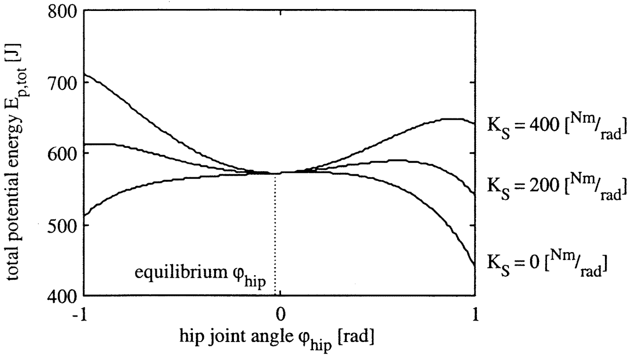
A. Hip Joint Stiffness versus Ankle Joint Stiffness

The effects and relation merits of applying hip-joint stiffness and ankle-joint stiffness were investigated by comparing both situations in simulations. For varying values of the stiffness K_S, the total potential energy was calculated. Stiffness values resulting in a stable equilibrium postures were found by checking for a local minimum in the energy function. Fig. 2(a) and (b) give the total potential energy $E_{p,\mathrm{tot}}$ as a function of the hip-joint angle φ_{hip} and ankle angle φ_1, respectively, for various values of hip-joint stiffness and ankle-joint stiffness, respectively.

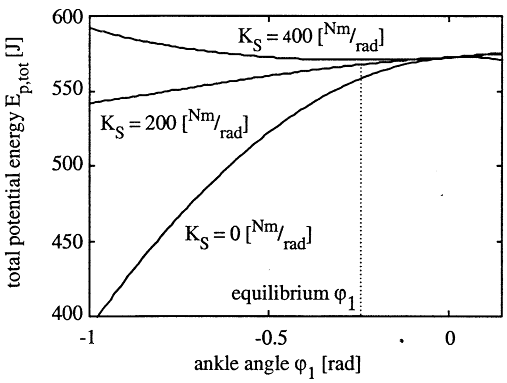
Fig. 2 expresses that increased hip-joint stiffnesses yield different results than increased ankle stiffness. When a hip-joint stiffness K_S of 200 N $\cdot$ m/rad is chosen a local minimum is predicted [Fig. 3(a)], representing an equilibrium. However, the same value of ankle-joint stiffness does not result in a local minimum, and thus in an unstable system [Fig. 2(b)]. Increasing the ankle-joint stiffness to 400 N $\cdot$ m/rad makes the system stable.

From this, it can be concluded that less stiffness is required at the hip joint than at the ankle joint to obtain stable stance. In addition, the large moments needed at the ankle may not be achievable as the feet may be lifted due to the limited moment arm at the ankle, especially in posterior direction.

From the parameters listed in Table I, all corresponding to values typical for humans, it can be calculated that for 400 N $\cdot$ m/rad of ankle stiffness, an ankle angle φ_1 over 0.4 rad will cause the ankle moment to become more than the ankle moment that can be generated under the foot. This maximal ankle angle of 0.4 rad is probably easily exceeded causing loss of balance, especially in dynamic situations where often overshoot is observed.



(a)



(b)

Fig. 2. (a) Total potential energy $E_{p,\mathrm{tot}}$ when hip-joint stiffnesses K_S of values 0, 200, and 400 N · m/rad are applied. (b) Total potential energy $E_{p,\mathrm{tot}}$ when ankle-joint stiffnesses K_S of values 0, 200, and 400 N · m/rad are applied.

For this reason, ankle-joint stiffness will not be very useful. The effect of hip-joint stiffness will be discussed for the remainder of this paper.

Results of the Static Analysis: First, the influence of changing hip-joint stiffness K_S and foot to crutch distance R_1 on the equilibrium postures, the equilibrium arm forces F_{arm} and on the hip-joint moments T_2 was studied. The results from these analyses are shown in Fig. 3.

Fig. 3(a) and (b) shows that increasing R_1 results in a reduced minimal stiffness required for stability. Furthermore, increasing K_S leads to a straighter hip joint in equilibrium. A larger K_S implies that the stabilizing hip-joint moment is larger, which causes the hip-joint angle in equilibrium to be closer to the reference value of 0 rad. The direct implication of a straightening hip joint is that the moment around the hip decreases as well [Fig. 3(c)]. Decreasing R_1 also straightens the hip joint [Fig. 3(a) and (b)] and decreases the arm force applied [Fig. 3(d)]. In addition, it is also clear that the equilibrium postures for different values of R_1 and K_S are all postures in which the hip-joint angle is smaller than zero. These postures represent the so-called C postures often observed in paraplegic patients when standing with crutches. All these postures will require between 200 and 250 N of armforce. The minimal stiffness K_S required for stability varies from 55.2 N · m/rad for $R_1 = 0.3$ m to 44.2 N · m/rad for $R_1 = 0.9$ N.

The offset hip-joint angle φ_0 was defined 0 rad in the results shown so far. The effects of changing φ_0 were also calculated. Therefore, φ_0 was varied between −0.2 (extension) and 0.2 rad (flexion) for the same range of K_S and R_1 as in the previous analysis results. The results are depicted in Fig. 4.

Fig. 4(b) shows that by changing the hip-joint reference angle, the equilibrium posture can be changed from an extension posture $(\varphi_0 < 0)$ to a flexion posture $(\varphi_0 > 0)$. However, changing the hip-joint reference angle also implies the need of increasing the stiffness, since the minimal stiffness needed to obtain stable postures increases changing the hip-joint reference angle. The arm forces decrease when the reference angle is chosen smaller than 0 rad.

Results of the Dynamical Analysis: For the dynamical analyses, not only hip-joint stiffness K_S but also hip-joint damping K_D was used to obtain sufficiently damped dynamic responses. Initially, the crutch-to-foot distance R_1, the stiffness K_S, and the offset hip-joint angle φ_0 were chosen at 0.3 m, 100 N · m/rad, and 0 rad, respectively. The crutch length d_3 was defined 1.35 m throughout this study. The damping K_D was calculated at 25.0 N · ms/rad. For this situation, the static analysis predicted an equilibrium hip-joint angle of −0.13 rad. For this configuration, the transient responses to three different magnitudes of moment disturbance in both directions were determined. The extension moment disturbances were chosen at −20, −25, and −30 N · m (unstable). The flexion moment disturbances were chosen 15, 20, and 25 N · m. The responses are shown in Fig. 5.

From these results, it can be seen that the −20 and −25 N · m of extension moment disturbance and the 15 and 20 N · m of flexion moment disturbance should lead to stable responses. Both −30- and 25-N · m disturbances resulted in instability. Further analysis showed that the maximal extension moment that lead to a stable response was −28.0 N · m. For flexion disturbances, the maximal flexion moment was 23.2 N · m. It can be seen from Fig. 5 that even for disturbances close to the maximum disturbance the responses are sufficiently damped as there is not much overshoot present while the equilibrium is restored within 3 s in all cases. The time responses for −30 and 25 N · m were unstable as both values are larger than the maximal values following from the dynamical analysis. In further analyses, the maximum disturbance values were determined for different system parameters.

For these analyses, the hip-joint stiffness K_S was varied as well as the crutch to foot distance R_1. The crutch length d_3 was chosen 1.35 m, and the offset hip-joint angle φ_0 was chosen 0 rad. For these values, the maximal flexion and extension disturbance moments were calculated before instability occurred and plotted in Fig. 6(a) and (b), respectively.

Fig. 6(a) and (b) shows that increasing the stiffness K_S or increasing R_1 makes the system more stable for both extension and flexion disturbance moments. If a static posture is achieved and balance is perturbed using a hip-moment pulse, more disturbance in extension direction is allowed for almost all static postures. This can be understood by looking at Fig. 3, which gives the static postures for the same parameters as in Fig. 6. Fig. 3 shows that the hip joint is in extension for all static postures. In these static postures, there is already a stabilizing hip-joint moment acting at the hip which is in equilibrium with the gravitational forces in the system. Applying a disturbance hip-joint moment in the extension direction will only increase this stabilizing moment, allowing

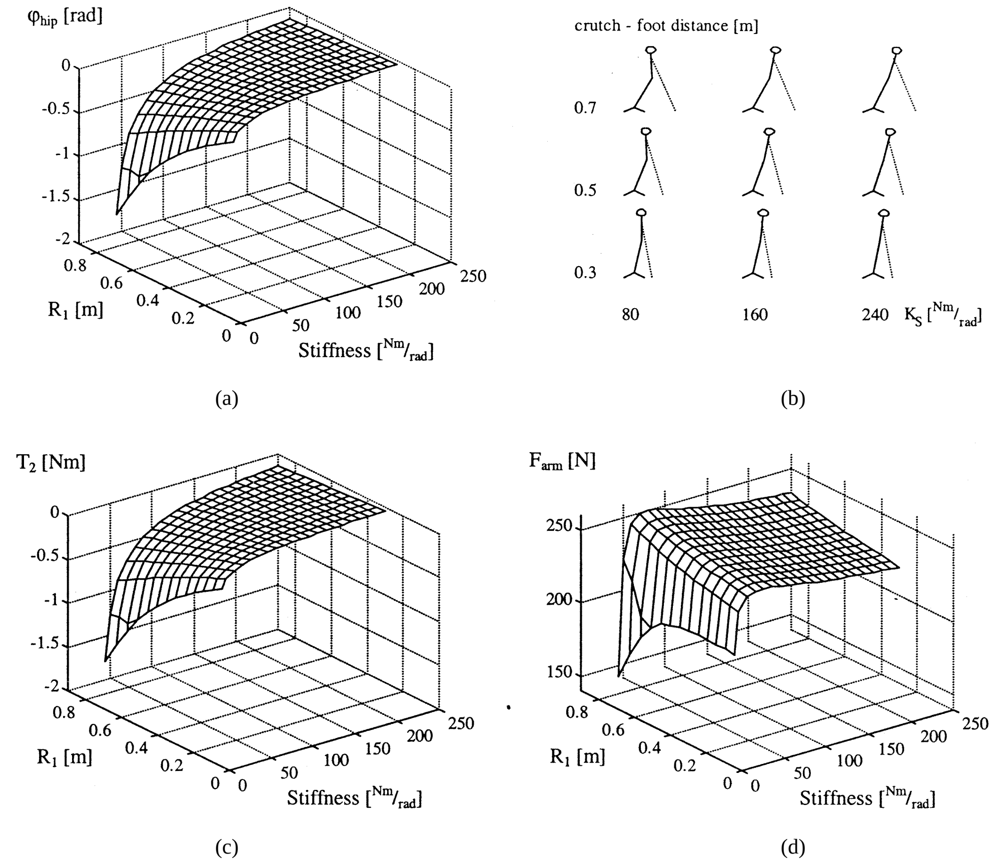


Fig. 3. Equilibrium values for: (a) hip-joint angle $\varphi_{\rm hip}$, (c) hip-joint moment T_2, and (d) arm force $F_{\rm arm}$ as a function of changing stiffness K_S and crutch-to-foot distance R_1. Values of K_S were 50, 60, ..., 250 N · m/rad while R_1 values were 0.3, 0.4, ..., 0.9 m. (b) Linkplots of some of the statical positions.

more disturbance to be applied in the extension direction. The postures for which more extension moment than flexion moment can be applied roughly coincide with the flatter part of Fig. 3(a). For these postures, generally, the hip-joint angle is smaller than -0.5 rad (that is, closer to 0 rad). If the equilibrium postures are more in extension, more flexion moment can be applied than extension moment.

Fig. 7 gives the results of the dynamical analysis for changing offset hip-joint angle φ_0 and stiffness K_S. Crutch-to-foot distance R_1 was chosen 0.3 m.

Fig. 7 shows that the maximal extension moment to increase markedly when the offset hip-joint angle is made positive (flexion), whereas the maximal flexion moment decreases. The inverse effect was observed when making the offset hip-joint angle negative. From Fig. 4, it can be seen that in the cases where the offset hip-joint angle is more positive the equilibrium postures show a flexed hip joint and small arm forces are applied. Lower arm forces are desirable, but the negative side effect is that these postures are not very stable for flexion disturbances. Conversely, disturbances in the extension direction will move the hip joint into a more stable direction. For negative offset hip-joint angles, the opposite effect holds. The equilibrium postures for these offset angles represent postures with decreased arm forces and extended hip joints. Fig. 4 demonstrates that the arm forces are not as low as when $\varphi_0 > 0$. Thus, more flexion disturbance can be applied than extension disturbance.

V. DISCUSSION

In literature, a large number of studies have used biomechanical modeling to analyze paraplegic standing and study the effects of applying closed loop control: A three degrees of freedom (DoF) skeletal model of paraplegic standing, also incorporating muscle dynamics, was constructed by Khang *et al.* [22], [23]. Muscle dynamics were incorporated into the model. An optimal output feedback controller was implemented, to minimize applied stimulation and, thus, muscle fatigue. Model results indicate that stable standing should be possible using this approach. There are numerous model based studies that have been verified by experiments which demonstrate that stabilization of the hip joint (anterior/posterior [24] and medial/lateral [25]) is a prerequisite to achieve successful stable paraplegic stance. In these studies, however, often control strategies were proposed *a priori* after which the controller's performance was determined, either using modeling or experiments. In our approach, however, we used a model and studied the biomechanics to determine what are the prerequisites to stable crutch supported standing. The actual way in which they are implemented must therefore be dealt with in a later study.

Modeling results were derived from a relatively simple biomechanical model [19] consisting of two linked segments representing the legs and the trunk. The choice of such a model is not a trivial matter. Biomechanical models tend to grow progressively more complicated when the number of

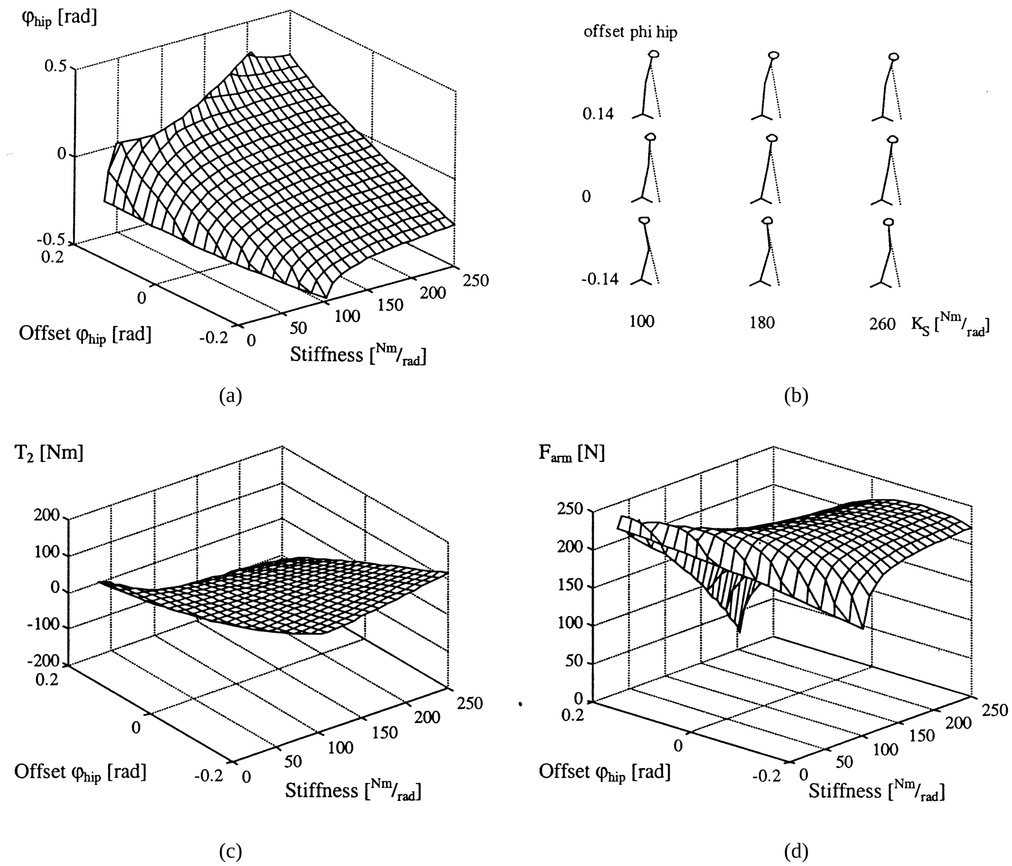


Fig. 4. Equilibrium values for: (a) hip-joint angle $\varphi_{\rm hip}$, (c) hip-joint moment T_2, and (d) arm force $F_{\rm arm}$ as a function of changing stiffness K_S and reference hip-joint angle $\varphi_{\rm href}$. Values of K_S were 50, 60, . . . , 250 N · m/rad, while $\varphi_{\rm href}$ values were -0.2, -0.18, . . . , 0.2 rad. (b) Linkplots of some of the statical positions.

segments is increased, demanding, likewise, increased numbers of parameters, which can be hard to identify from literature or direct measurements and which can vary considerably between different individuals. On the other hand, it has been shown by Barin [26] that in order to properly describe the dynamics of paraplegic standing, a single inverted pendulum model is not appropriate, needing at least two different segments to be included in the model.

The two-link segment model used in our study is relatively simple and there are many deviations from reality. Yet, the model is thought to describe several of the essential features of crutch supported paraplegic standing. Some of the deviations from reality are mentioned in the following.

In the model, the arm forces were defined as the constraint force acting on the shoulder keeping the tip of the crutches and the shoulder at a fixed distance d_3. In this situation, the influence of arm forces exerted on crutches is considered static so the shoulders do not actively participate in balance. In practice, the shoulder (and, thus, the upper part of the trunk) will move and moments will be generated by the upper trunk. This needs to be considered when comparing the modeling results with experimental findings.

Since only two segments are included in the model, the trunk and shoulders were assumed rigid. In practical situations, however, this will not be the case. We expect, therefore, that under dynamic conditions, the model will be less accurate than under static conditions. Under dynamic conditions a considerable part of the disturbances applied may be compensated or even dissipated by movements of the upper trunk and shoulders. Probably, this effect will increase when applying higher values of hip-joint stiffness. In static situations however, there obviously is no movement in the trunk and shoulder, so the model is more likely to provide realistic results.

The model we used was supposed to be simple, yet to provide sufficient insight in the biomechanics of crutch supported paraplegic stance in clinical settings. We aim to extend our work with experiments on paraplegic subjects who will be wearing orthoses for medial-lateral support. In those clinical settings, no problems of instability in the medial-laterial plane are expected. If, however, crutch supported standing is done with only one crutch, medial-laterial instability may occur in the form of axial rotation. In future modeling work, this could be further analyzed, e.g., by introducing an additional degree of freedom of the hip and running a new set of simulations with an extended model.

Even our relatively simple two-segment model predicts nontrivial properties. The potential energy function leads to unstability in standing when realistic parameters are used. The potential energy in the system depends on geometrical parameters such as the crutch length d_3, the crutch-to-foot distance R_1, the hip-joint stiffness K_S, and the offset hip-joint angle φ_0 (hip-joint angle for which no torque is generated by the stiffness). These parameters will affect the occurring equilibrium postures as well. The occurring of instability is not self evident:

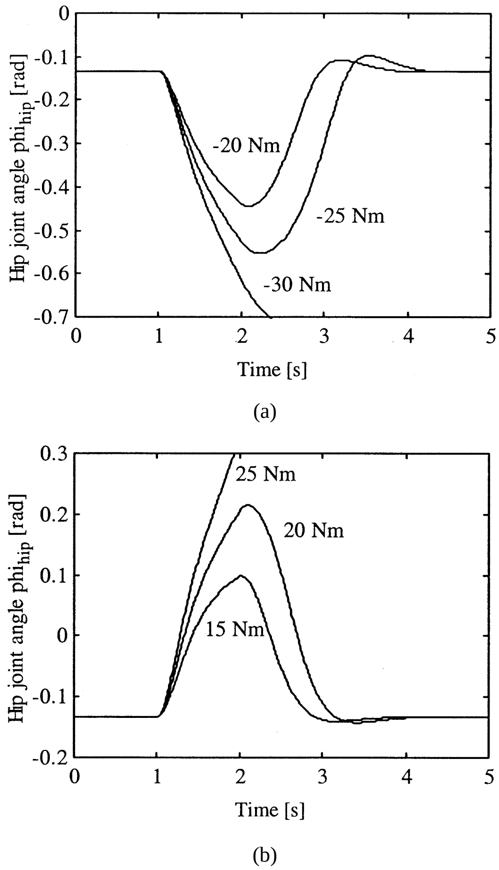


Fig. 5. Transient responses of simulated balance perturbing trials. Pulse shaped, 1 s duration hip-joint moment disturbances of different amplitudes were applied at $t = 1$ s in (a) extension and (b) flexion direction.

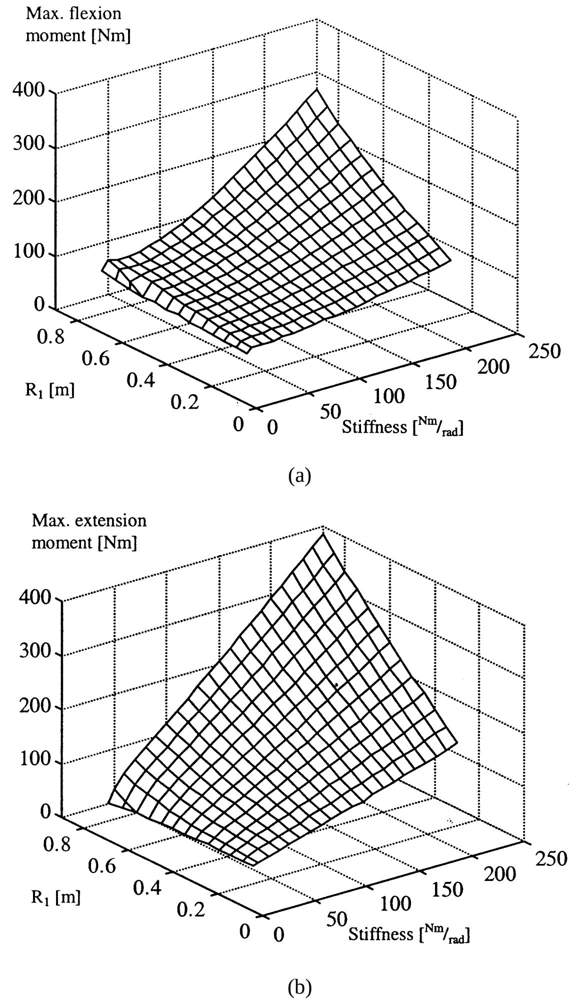


Fig. 6. Maximum disturbance moments that can be applied in (a) flexion and (b) extension direction for changing stiffness K_S and foot to crutch distance R_1. Values of K_S were $50, 60, \ldots, 250$ N $\cdot$ m/rad while R_1 values were $0.3, 0.35, \ldots, 0.9$ m. Crutch length d_3 was 1.35 m.

When the ankle angle φ_1 is changed, the two segments move in opposite directions and, therefore, change their potential energy in opposite directions. This leads to a flattening of the potential energy curve seen in Fig. 2. Our study could be viewed as an analysis of how this curve can be modified as to achieve an optimally stable curve.

Matjacic *et al.* [16], [17] used increased artificial ankle-joint stiffness to do this and found that a paraplegic subject was able to maintain upright standing when using sufficiently high stiffness around the ankle. In their experimental setup, the subject was placed on a platform that was controlled by an electromotor generating the required stiffness. The feet were attached rigidly to this platform and the subject used trunk efforts to balance. As a result of this configuration, large stiffnesses around the ankle could be generated, because the danger of heel- or toelifting was not present. In our application, we assume crutch supported paraplegic standing where high stiffnesses around the ankle cannot be applied because in our setup heel- or toelifting is possible. Therefore, hip-joint stiffness was applied instead of ankle-joint stiffness.

Modeling results predict that it should be possible to balance a crutch supported paraplegic subject by using additional hip-joint stiffness. The equilibrium postures that were found when taking the offset hip-joint angle $\varphi_0 = 0$ rad were all C postures (i.e., postures in which the hip joint was in extension).

By increasing the offset hip-joint angle φ_0 to a small flexion angle, it is possible to obtain more erect stable postures. This will also lead to decreased arm forces exerted on the crutches. Therefore, it seems that changing the offset hip-joint angle φ_0 is a useful way of changing the static posture without causing the arm forces to increase. In dynamic situations however, less flexion disturbance can be applied before instability occurs.

In the Introduction, three requirements of crutch supported standing were formulated: stability, minimized arm forces, and maximized freedom of movement. Regarding these requirements the following can be said: The minimal stiffness needed for stability is dependent on geometrical parameters, but for a hip-joint stiffness above 60 N $\cdot$ m/rad, stable postures were possible in all simulations in which the reference hip-joint angle was taken zero. Making the reference hip-joint angle negative, a higher stiffness has to be used to obtain stable postures. For values of hip-joint stiffness above 80 N $\cdot$ m/rad, the arm forces were typically around 200 N. When changing the reference hip-joint angle, the equilibrium postures (C postures) can be changed into postures with straighter hip joints. This will result in decreased arm forces and a decreased ability to compensate for disturbance in the posterior direction.

Conclusions about the patient's ability to handle objects are difficult to make based on a simulation study only. With a stiffer

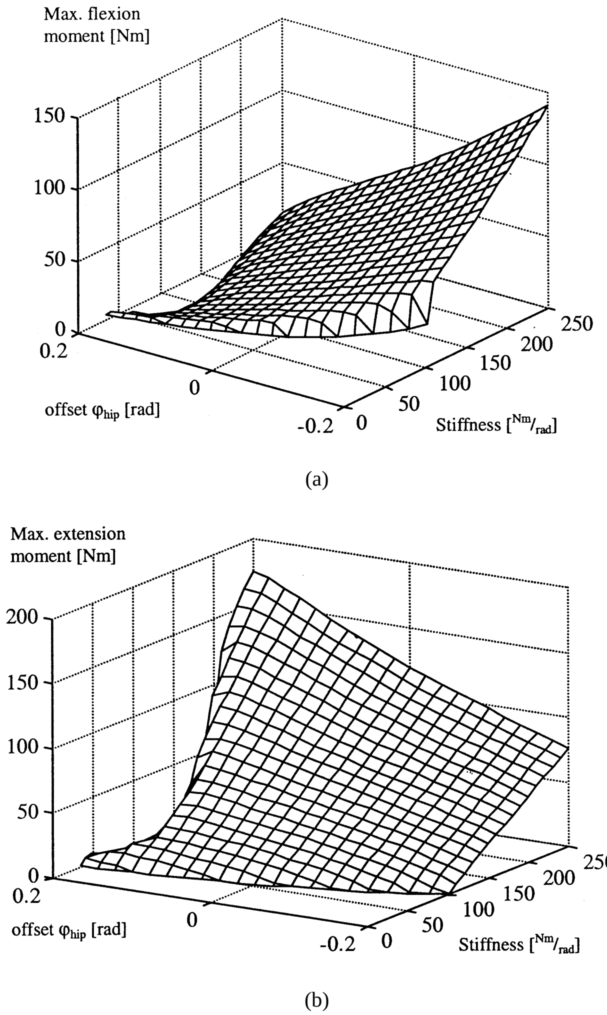


Fig. 7. Maximum disturbance moments that can be applied in (a) flexion and (b) extension direction for changing stiffness K_S and reference hip-joint angle φ_{href}. Values of K_S were 50, 60, . . . , 250 N · m/rad, while φ_{href} values were -0.2, -0.18, . . . , 0.2 rad. Crutch length d_3 was 1.35 m.

hip joint, the freedom of movement will evidently decrease, because more effort is needed to flex or extend the hip joint. A stiffer hip joint will result in an equilibrium posture with a straighter hip joint. This posture may be altered to obtain a C posture by changing the offset hip-joint angle φ_0. However, the allowable disturbances will decrease in this situation. Future experiments will have to make clear what practical tasks a patient can perform when standing, depending on applied hip-joint stiffness.

The application of hip-joint stiffness and hip-joint damping may be a useful tool in the control of crutch supported paraplegic stance. Changing various parameters gave considerable changes in static postures and dynamic behavior. Changing the reference hip-joint angle in such a way that an equilibrium posture is obtained with a nearly aligned hip-joint results in decreases in the arm forces exerted on the crutches. In addition, the simulation results suggest that static postures can not be obtained when no additional hip-joint stiffness is applied. This is in accordance with our preliminary experimental findings [19].

Estimation of the damping using a small signal analysis around the equilibrium posture gave a sufficiently damped system when applying disturbances to the (nonlinear) system and controlling balance with a linear stiffness. The damping parameter K_D was included in the model to simulate the effect of natural and artificial damping present under experimental conditions. To prevent unrealistic oscillations in the responses to disturbances, a small signal analysis for the highly nonlinear system was used to provide a reasonable starting point of the damping K_D.

Whether nonlinear stiffness will improve performance has yet to be determined. Whether a nonlinear controller matched to this system would allow for a static equilibrium "range" instead of a single static equilibrium posture is questionable. In this situation, movements within the equilibrium range would be possible without disturbing equilibrium. This would allow a paraplegic subject standing with crutches to move voluntarily within a certain range and still be in equilibrium. This would be desirable should the subject want to handle or grasp objects. One of the problems is, however, that the required stiffness is highly dependent of geometrical parameters like the crutch to foot distance R_1. Continuous online measurement of these parameters will, therefore, be necessary to account for their influences. It can be argued whether a very complex nonlinear controller needing much sensory information is preferrable to a simple and robust controller, being a spring.

The modeling results must be compared with experimental data. Actually, the model gives many direct ways of validating the predictions. Hip-joint stiffness can be implemented by placing spring over a hip–knee–ankle–foot–orthosis' (HKAFO) hip joint. The stiffness can be varied by varying the diameter of the springs. More flexible control can be accomplished by electrical stimulation of hip-joint muscles in a closed loop.

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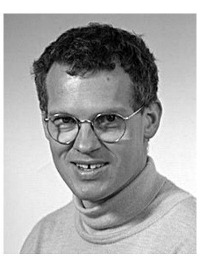

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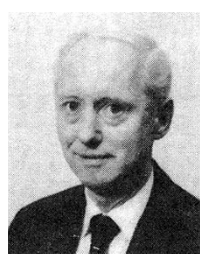

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