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Joint torques during sit-to-stand in healthy subjects and people with Parkinson's disease

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Abstract

Objectives. To compare lower limb joint torques during sit-to-stand in normal elderly subjects and people with Parkinson's disease, using a developed biomechanical model simulating all phases of sit-to-stand.

Design. A cross-sectional study utilizing a Parkinsonian and a control group.

Background. Subjects with Parkinson's disease were observed to experience difficulty in performing sit-to-stand. The developed model was used to calculate the lower limb joint torques in normal elderly subjects and subjects with Parkinson's disease, to delineate possible causes underlying difficulties in initiating sit-to-stand task.

Methods. Six normal elderly subjects and seven age-matched subjects with Parkinson's disease performed five sit-to-stand trials at their self-selected speed. Anthropometric data, two-dimensional kinematic and foot-ground and thigh-chair reactive forces were used to calculate, via inverse dynamics, the joint torques during sit-to-stand in both before and after seat-off phases. The difference between the control and Parkinson's disease group was analysed using independent *t*-tests.

Results. Both control and Parkinson's disease groups had a similar joint kinematic pattern, although the Parkinson's disease group demonstrated a slower angular displacement. The latter subjects produced significantly smaller normalized hip flexion torque and presented a slower torque build-up rate than the able-bodied subjects ($P < 0.05$).

Conclusion. Slowness of sit-to-stand in people with Parkinson's disease could be due to a reduced hip flexion joint torque and a prolonged rate of torque production.

Keywords: Sit-to-stand; Ground reaction force; Functional evaluation; Joint torque; Parkinson's disease

1. Introduction

Rising from a chair involves the transition from a stable seated position to a relatively unstable upright stance, and requires coordinated contractions of the muscles of the lower extremities and trunk. Specifically, subjects with Parkinson's disease (PD) have been reported to experience difficulties in rising from a seated position, and demonstrate impairments in the ability to control sequential and/or coordinated movements of the joints (Nikfekr et al., 2002; Seidler et al., 2001; Serrien et al., 2000; Swinnen et al., 2000). PD subjects are also known to experience problems in movement initiation. This can be largely attributed to muscle weakness and reduced abilities to generate rapid muscle contractions (Corcos et al., 1996; Kakinuma et al., 1998). Despite the importance of sit-to-stand (STS) movement in daily-life activity, there are virtually no reports on the lower limb dynamics during STS in PD subjects. Consequently, it remains unsettled whether motor deficits observed in PD subjects during chair-rise are related to the lower limbs' torques, and whether muscle weakness and rate of force generation contribute to the reduced ability to perform STS activity in PD subjects.

Previous works studying the biomechanical aspects of STS activity included data on the reactive forces, kinematics of the lower limb segments, kinetics of the center of mass (CoM) and joint torques. The populations studied were able-bodied subjects, elderly subjects and subjects with neurological pathologies

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(Nikfekr et al., 2002; Gross et al., 1988; Lundin et al., 1995; Pai et al., 1994; Rodosky et al., 1989; Roebroeck et al., 1994; Seedhom and Terayama, 1976; Shepherd and Gentile, 1994; Wretenberg and Arborelius, 1994; Scarborough et al., 1999; Schenkman et al., 1996). Calculation of the joint torques, as carried out on able-bodied subjects, revealed an uncertainty due to the unknown reactive forces between the thighs and the chair (Gross et al., 1988; Lundin et al., 1995; Pai et al., 1994; Rodosky et al., 1989; Roebroeck et al., 1994; Seedhom and Terayama, 1976; Shepherd and Gentile, 1994). Thus, the lower limb joint torques could only be reliably derived from the instant of seat-off onwards (Rodosky et al., 1989; Roebroeck et al., 1994). Neglecting the forces acting on the thighs would result in under-estimating the net joint torques at the hip and knee during the forward flexion phase of STS, when the thighs are still in contact with the chair.

Another aspect in which no previous reports were found is the errors in estimating the joint torques. These errors typically involve several sources of uncertainties related to joint positioning, segments' anthropometrical properties, deformity of the trunk during movement, synchronization errors between force plate and motion analysis system and digitization errors of the joint positions (Kingma et al., 1996; Kuo, 1998; Levin et al., 1998). However, if the system of equations in a linked segment model is over-estimated, it may become possible to use the redundancies to test the reliability of the results on the basis of full consistency with the boundary conditions, by using a least-square estimation approach (Kuo, 1998;Levin et al., 1998).

The primary aim of the present study was to compare the torque production capacity in the lower limb joints between PD and age-matched able-bodied subjects, before and after the seat-off phase of STS movement. Kinematic data and force-plate measurements of the foot-ground and thigh-chair reaction forces provided redundant input data, allowing us to test the reliability of the results. Knowledge of the lower limb joint torques prior to the seat-off phase is essential to understand possible motor disorders underlying chair-rise failure in patients with PD. The torque profiles provide information on muscle strength, force build-up rate and intermuscular coordination across the joints. These factors are considered important and served in the past to determine and/or parameterize dynamic stability in human subjects, and were shown to provide an indicator of deterioration in the motor performance of subjects with motor disorders (Seidler et al., 2001; Serrien et al., 2000; Scarborough et al., 1999; Schenkman et al., 1996). Previous studies have also shown that elderly subjects tend to adopt exaggerated flexion strategies to compensate for weakness of the quadriceps muscles in controlling range of motion and speed of trunk flexion (Scarborough et al., 1999). Accordingly, it is expected that the torque profiles in the knee and hip during the seat-off phase can provide information also about the control strategies used by PD subjects to achieve chair-rise movements.

2. Experimental procedure

2.1. Subjects

Thirteen elderly adults, including six healthy subjects and seven patients with the diagnosis of idiopathic PD (Stages 2.5–3 of Hoehn and Yahr Staging Scale, 1967) took part in this study. The able-bodied subjects had a mean age of 69.2 years (SD, 4.0 year, range 63–72 years); their average mass was 59.2 kg (SD, 11.4 kg, range 43–75 kg); and their average height was 160 cm (SD, 8.7 cm, range 147–169 cm). The patients with PD had a mean age of 66.9 years (SD, 7.9 years, range 53–73 years); their average mass was 52.4 kg (SD, 8.3 kg, range 41–67 kg) and their average height was 156 cm (SD, 8.8 cm, range 138–164 cm). They were all stable on anti-Parkinsonian medication, and could stand up from a chair independently upon verbal instruction without the use of hands. None of the subjects had any history of orthopedic, arthritic or heart diseases. All subjects gave their informed consent and all the experimental work was carried out with the approval of the local ethical committee.

2.2. Task and protocol

The subjects were positioned on an armless, adjustable chair that was mounted on the surface of a strain gauge force-plate (platform B, Fig. 1). The height of the chair was such that, in all subjects, the knees were flexed at approximately 90° when the feet were positioned on the surface of a second force plate (platform A). From this initial position, all subjects were able to rise independently, without further flexing their knees or taking a step and without unfolding their arms. The requirement for no hand contact with the chair was necessary to allow us to monitor the full range of torque production capacity in the ankle, knee and hip. The force plates used (AMTI, Advanced Medical Technology Incorporated, Watertown, USA) were aligned longitudinally 8 cm apart from each other (Fig. 1). The separation between the feet was equal to that of the shoulder width. The position of the feet on the surface of the force plate was similar for all subjects, and was symmetrical in relation to the longitudinal axis of the platform. Reflective markers, 2.5 cm diameter, were placed over the lateral left-hand side of the body, on the head of the fifth metatarsal, lateral malleolus, femoral epicondyle, greater trochanter, acromion process, and in front of the ear canal (Fig. 1).

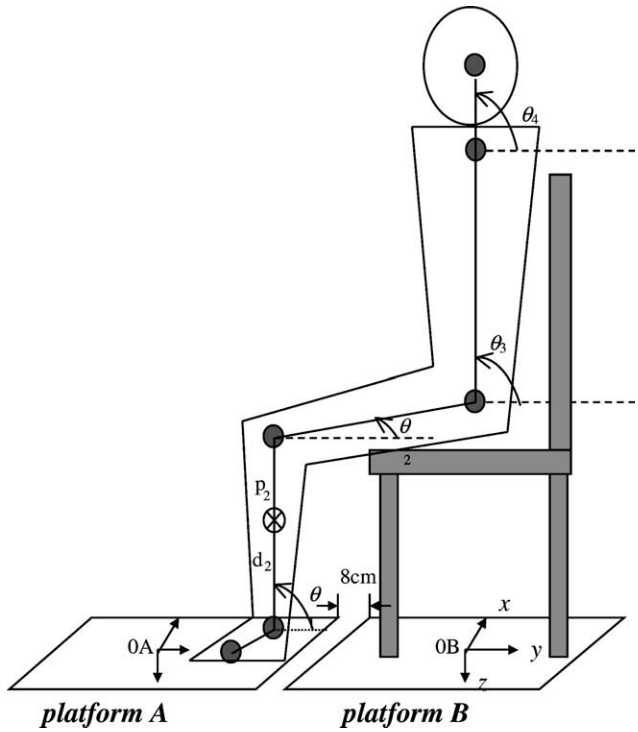

Fig. 1. Schematic description of the four-joint, five-segments presentation of the human body, including the feet, lower legs (shanks), upper legs (thighs), trunk and head. Definitions for the joint angles were as follows: θ_1—ankle, θ_2—knee, θ_3—hip and θ_4—neck. $\boldsymbol{p}_2$ and $\boldsymbol{d}_2$ are the position vectors connecting the CoM ($\otimes$) to the proximal and distal joints of the shank segment.

Subjects sat with the trunk erect and the arms folded over their chest. They were asked to wait for a verbal signal before initiating the STS movement. Following the verbal signal, the subject rose from the chair at a self-selected speed and maintained a stand-still upright position. Each subject performed eight trials of the STS task: the first three served as practice trials and in the remaining five, force plate and video data were monitored. Zero-setting of the force readings from the unloaded platforms were made prior to data recording, to correct for long-term drift, DC offsets and the weight of the chair on platform B.

2.3. Data collection

Data collection was initiated 3 s prior to the verbal signal and lasted 9 s for each trial. Reaction forces were collected from the two force platforms at 100 samples per second, and were digitally filtered with a zero-phase lag bi-directional (forward and backward) fourth-order low-pass Butterworth filter at a cut-off frequency of 5 Hz (Kuo, 1998; Kingma et al., 1995). The motion of the reflective markers was recorded using a video camera operating at 50 frames per second. The camera was placed perpendicular to the plane of motion at a distance of 3.5 m, and opposite to the left sagittal side of the subject. To allow synchronization between kinematics and force plate data, a signal from a light emitting diode was indicated on the videotape over 10 picture frames, 3 s from the onset of the force plate data collection. Kinematic data were extracted from videotape recordings, using a two-dimensional video-based registration system (PEAK, Performance Technologies Inc., Englewood, USA). The horizontal and vertical positions of each reflective marker were tracked from each frame, by using the PEAK built-in data acquisition program for automatic digitization. Following calibration, the absolute coordinates were calculated and low-pass filtered with a zero-phase lag, bi-directional, fourth-order Butterworth filter at a cut-off frequency of 5 Hz.

2.4. Data analysis

A linked segment model of the human body was used to calculate the lower limb joint torques during all the phases of STS movement. An explicit description of the model is presented in Appendix A. Each segment provided a set of dynamic equations, relating together gravitational forces, and inertial and reactive forces and moments. The measured force plate data and the recorded motion of the five segments (feet, lower legs, upper legs, trunk and head) were fed into the model solution algorithm presented in Fig. 2, from which the instantaneous net moments about the ankle, knee, hip and neck joints were calculated. The model's solutions in the distal segments were then compared with those actually measured. Differences between model and measured values were calculated and minimized, using the iterative algorithm presented in Fig. 2, to re-estimate the joint positioning and point of application of the chair-thighs reaction force. Curves of the net joint moments (joint torques) were calculated for each segment and normalized with respect to body-weight times height ($\mathrm{N\,m\,Kg^{-1}\,m^{-1}}$). Peak values of the neck, hip and knee extension and flexion torques and ankle plantar- and dorsiflexion torques were extracted from the moment curves. Next, the time-to-peak torque was obtained as the time elapsed between STS onset to the point of peak joint torque, from which the rate of torque build-up was derived. Values obtained were compared between the two subject groups.

2.5. Statistics

Differences between the control group and the patients with PD for the peak reaction forces, peak joint angles, peak torques and the rate of torque production were tested with student's t-tests, and the level of significance was determined at $P < 0.05$.

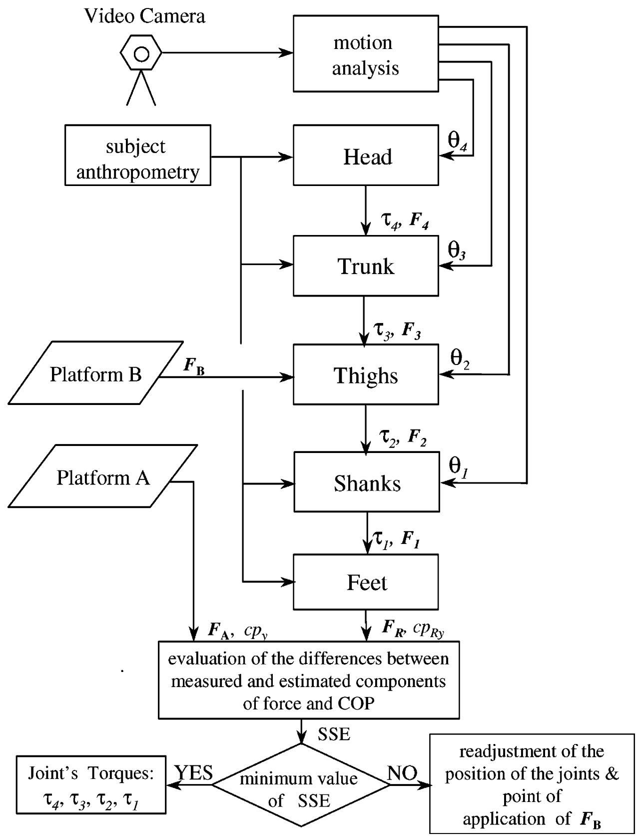

Fig. 2. Block diagram of the model. Kinematics of the segment were estimated directly from video monitoring of the body segments. Forces and moment acting on the model segments were resolved by means of backward dynamics calculation. The reaction forces between the thigh and the chair and their approximated point of application with respect to the knee joint were taken on the basis of the measurements from platform B (F_B). Differences between the model (F_R and cp_R) and measured (F_A, cp_y) components of foot-ground reaction forces and the point of application were minimized by re-adjustment the positioning of the joints and the approximated positioning of the point of application of F_B. The moments at the joints were estimated afterward. The explicit kinematics and dynamics calculations are summarized in Appendix A.

3. Results

3.1. Force-plate data during chair-rise in able-bodied and PD subjects

Fig. 3 illustrates foot-ground (F_A) and seat-thigh (F_B) reaction forces during STS movement of an able-bodied and a PD subject. The first transitory portion of the anteroposterior components of the foot-ground and seat-thigh reaction forces (with the negative peak) indicates the forward acceleration phase of the STS movement. The forward acceleration phase was followed by a steep increase in the vertical component of F_A (dashed curve), and a steep decrease in the vertical component of F_B (solid curve). The next transitory portion with positive peaks in the anteroposterior and vertical components of F_A indicates the upward acceleration (rising) phase of the STS movement. Seat-off was determined as the instant when the components of the seat-thigh contact force (F_B) became zero. Relative to the able-bodied subjects, the aforementioned transitions in foot-ground and seat-thigh contact forces were less extensive among PD subjects. In particular, we observed longer time intervals between the onset of movement and seat-off in patients with PD than the able-bodied subjects (mean duration: 1.4 s, SD, 0.4 s versus 0.8 s, SD, 0.2 s, $P = 0.015$), and lower forward and upward acceleration forces as compared with those observed for the able-bodied subjects. A quantitative analysis of the difference between the two populations for the reaction forces is presented in Table 1.

3.2. Joint kinematics during chair-rise in able-bodied and PD subjects

Patterns of angular displacements of the ankle, knee, hip and neck joints are illustrated in Fig. 4 for an able-bodied and a PD subject. In the able-bodied subject, the hip angle (θ_3) typically flexed from about 80° at the beginning of the forward acceleration to about 120° at the instant of seat-off, then extended to about 90° during the rising phase. The knee joint angle (θ_2) remained constant during the forward acceleration phase, then started to extend at seat-off until full extension at the end of the rising phase. When compared with able-bodied subjects, patients achieved similar peak joint angles as reported in Fig. 4 and Table 2. However, these patients took a longer time to reach peak neck extension, hip flexion and ankle dorsiflexion during the forward acceleration phase, and peak hip and knee extension during the upward acceleration phase of STS (Table 2).

3.3. Joint dynamics during chair-rise in able-bodied and PD subjects

Curves of the torques acting in the joints are presented in Fig. 5 for an able-bodied and a PD subject. The estimated torque about each joint was normalized with respect to body-mass times height, to allow for comparison between subjects of different body size. Typically, the normalized extension/flexion torques about each of the above joints leveled below 0.1 $\mathrm{N\,m\,Kg^{-1}\,m^{-1}}$ prior to the onset of movement. Initiation of the STS movement was accompanied by an increase in the hip flexion and ankle dorsiflexion torques at the ankle, as the center of pressure (CoP) of the foot-ground force moved posteriorly toward the heel. Peak hip flexion torque was reached soon after the onset of forward acceleration. The hip and the knee extension torques increased simultaneously during the forward accelera-

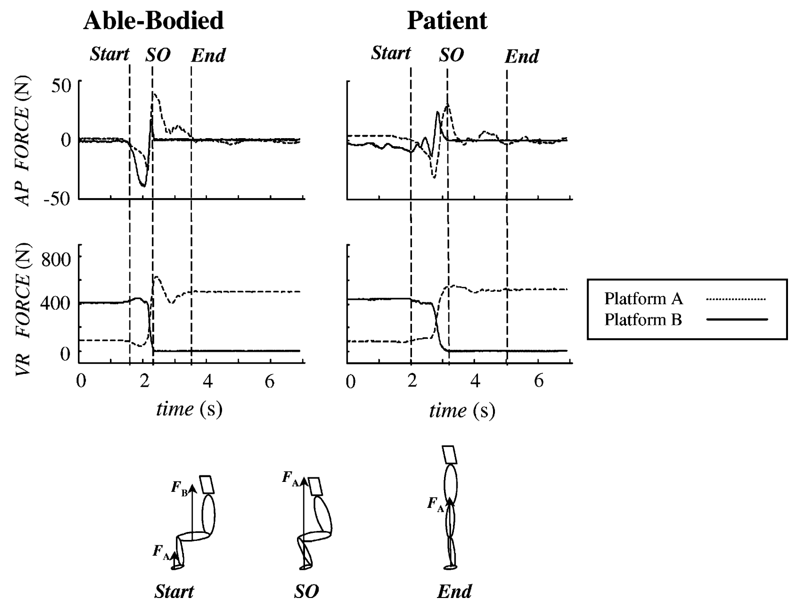

Fig. 3. An example for the anteroposterior (AP) and vertical (VR) reaction forces between the thigh and chair (platform B) and between feet and ground (platform A). The vertical lines indicate the limits between STS phases (Start = start of movement, End = end of movement and SO = the instant of seat-off).

Table 1
Peak anteroposterior and vertical reaction forces between feet and ground (platform A), and between the thigh and chair (platform B) for the able-bodied subjects and patients with PD

	Able-bodied ($n = 6$)	Patients with PD ($n = 7$)	P level
Foot-ground reaction force (N)—platform A			
Anteroposterior direction	47.6 (12.9)	27.3 (10.8)	0.010[a]
Vertical direction	610.6 (62.4)	475.7 (51.9)	0.030[a]
Seat-thigh reaction force (N)–-platform B			
Anteroposterior direction	62.7 (14.9)	25.8 (11.8)	0.013[a]
Vertical direction	39.0 (18.9)	17.1 (18.5)	0.059

Values shown are means (standard deviations).
[a] Significant difference between the two groups at $P < 0.05$.

tion phase, and reached their peak values very close to the instant of seat-off. Afterward, the extension torques at the hip and the knee joints started to descend, while the ankle torque reversed direction as the CoP of the foot-ground forces moved anteriorly to the ankle joint at the end of the movement. The pattern of the joint torques was similar in the PD subjects, except that the build-up rate of the torque in the hip, knee and ankle joints was distinctively slower when compared with that of the able-bodied subjects. The neck torque in both groups of subjects was considered negligible. Table 3 summarizes the mean and standard deviation values of the estimated normalized peak torques at the neck, hip, knee and ankle joint for the able-bodied and PD subjects. Typically, the levels of the normalized peak torque

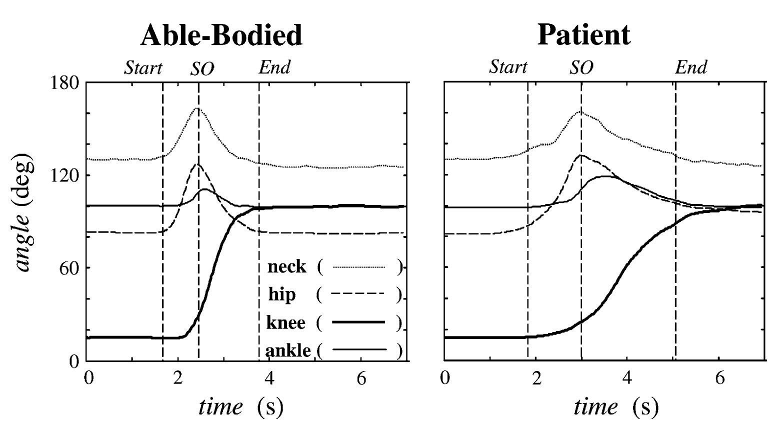

Fig. 4. Angular displacements about the neck (dotted curve), hip (dashed curve), knee (bolded curve) and ankle joints (solid curve) for an able-bodied subject and a patient with PD during STS transfer. The vertical lines indicate the boundaries between the motion phases (Start = start of movement, End = end of movement and SO = the instant of seat-off).

Table 2
Peak joint angles and time-to-peak angles from STS onset for neck, hip, knee and ankle joints for the able-bodied subjects and patients with PD

	Able-bodied ($n = 6$)	Patients with PD ($n = 7$)	P level
Angle(°)			
Neck extension	18.9 (11.2)	18.7 (8.0)	0.962
Hip flexion	29.7 (6.1)	34.4 (6.6)	0.212
Hip extension	87.1 (8.7)	83.2 (9.1)	0.397
Knee extension	78.1 (9.9)	73.7 (7.4)	0.382
Ankle dorsiflexion	10.7 (4.8)	13.9 (4.8)	0.254
Time-to-peak joint angle (s)			
Neck extension	0.18 (0.31)	0.95 (0.87)	0.416
Hip flexion	0.75 (0.22)	1.21 (0.34)	0.017[a]
Hip extension	1.43 (0.80)	2.30 (1.98)	0.060
Knee extension	1.37 (0.79)	2.12 (2.06)	0.063
Ankle dorsiflexion	1.17 (0.41)	1.54 (0.37)	0.115

Values shown are means (standard deviations).
[a] Significant difference between the two groups at $P < 0.05$.

obtained for the able-bodied group were of the same magnitude as those of the PD subjects. The only exception observed was for the peak values of the hip flexion torque in the able-bodied subjects, which was significantly higher than those of the PD subjects ($P = 0.023$). Table 3 summarizes the rate of torque build-up in the neck, hip, knee and ankle joints in the able-bodied and PD subjects. The results show that PD subjects had a significant tardiness in their capacity to generate hip flexion and extension, knee extension and ankle dorsiflexion torques, when compared with the able-bodied subjects (all, $P < 0.05$). The build-up rate of hip flexion torque in able-bodied was approximately fourfold faster than in PD subjects. The build-up rate of the hip and knee extension torques and the ankle dorsiflexion torque was approximately 150% faster in the able-bodied than in PD subjects.

4. Discussion

4.1. Dynamic performance during chair-rise

In both the able-bodied and PD subject groups, the peak hip flexion torque was reached soon after the onset of forward acceleration. The hip and knee extension and the ankle dorsiflexion torques reached their peak values very close to the instant of seat-off, whereas the peak ankle plantarflexion torque was reached towards the end of the STS task. This sequence is consistent with the findings previously reported (Gross et al., 1988; Lundin et al., 1995; Pai et al., 1994; Rodosky et al., 1989; Roebroeck et al., 1994; Arborelius et al., 1992; Bajd et al., 1982; Carr and Gentile, 1994; Doorenbosch et al., 1994; Fleckenstein et al., 1988; Pai and Rogers, 1991; Schultz et al., 1992). Regarding the magnitude of joint torques, both the able-bodied and PD subject groups generated the highest joint torque during knee extension, followed by hip extension, ankle plantarflexion, ankle dorsiflexion and hip flexion. This trend is consistent with previously reported dynamic calculations. However, in these studies, only body kinematics and foot-ground reaction forces were used as model inputs for the dynamic calculations of the joint torques, whereas the chair-thigh contact forces were not taken into account (Gross et al., 1988; Lundin et al., 1995; Pai et al., 1994; Rodosky et al., 1989; Roebroeck et al., 1994; Seedhom and Terayama, 1976; Shepherd and Gentile, 1994). Failing to take the chair-thigh contact forces may result in an under-estimation of the knee and ankle torques during the forward acceleration phase prior to seat-off.

When able-bodied subjects performed STS, the peak hip extension torque found in the present study was 0.91 $\mathrm{N\,m\,kg^{-1}\,m^{-1}}$. This was slightly larger than that reported in the previous studies, being 0.6–0.82 $\mathrm{N\,m\,kg^{-1}\,m^{-1}}$ for elderly able-bodied subjects (Gross

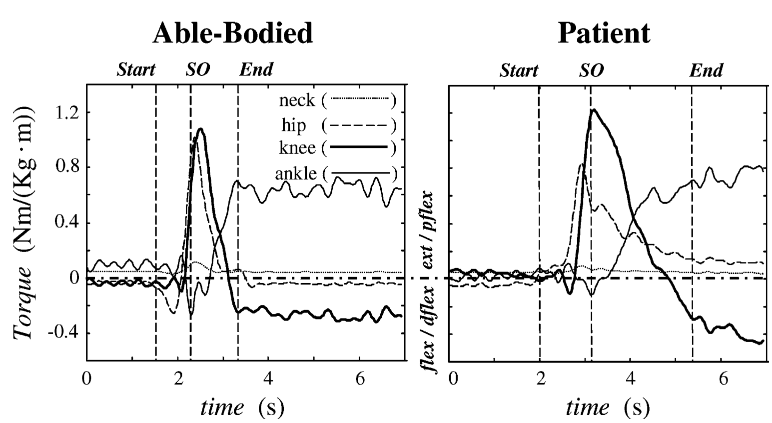

Fig. 5. Joint torques about the neck (dotted curve), hip (dashed curve), knee (bolded curve) and ankle (solid curve) for an able-bodied subject and a patient with PD during STS transfer. Torques are normalized to body-mass times height. The vertical lines indicate the boundaries between the motion phases (Start = start of movement, End = end of movement and SO = the instant of seat-off).

Table 3
Estimated net peak torques and rate of torque build-up in the neck, hip, knee and ankle joint for the able-bodied subjects and patients with PD

	Able-bodied ($n = 6$)	Patients with PD ($n = 7$)	P level
Net peak joint torque ($N\,m\,kg^{-1}\,m^{-1}$)			
Neck flexion	−0.012 (0.011)	−0.027 (0.012)	0.062
Neck extension	0.086 (0.026)	0.096 (0.008)	0.443
Hip flexion	0.201 (0.061)	0.089 (0.072)	0.023[a]
Hip extension	0.914 (0.173)	0.868 (0.116)	0.611
Knee flexion	0.207 (0.120)	0.234 (0.141)	0.746
Knee extension	1.170 (0.268)	1.066 (0.194)	0.470
Ankle dorsiflexion	0.378 (0.192)	0.228 (0.164)	0.194
Ankle plantarflexion	0.639 (0.167)	0.683 (0.113)	0.543
Rate of torque build-up ($(N\,m\,kg^{-1}\,m^{-1})\,s^{-1}$)			
Neck flexion	0.137 (0.024)	0.097 (0.012)	0.188
Neck extension	0.003 (0.004)	0.003 (0.006)	0.943
Hip flexion	1.033 (0.733)	0.260 (0.379)	0.033[a]
Hip extension	1.870 (0.603)	1.149 (0.394)	0.025[a]
Knee flexion	0.192 (0.009)	0.180 (0.149)	0.866
Knee extension	1.989 (0.451)	1.336 (0.447)	0.024[a]
Ankle dorsiflexion	0.476 (0.132)	0.325 (0.009)	0.032[a]
Ankle plantarflexion	0.634 (0.340)	0.326 (0.293)	0.106

Values shown are means (standard deviations).
[a] Significant difference between the two groups at $P < 0.05$.

et al., 1988; Schultz et al., 1992). The estimated peak knee extension torque reported here for the able-bodied group was 1.17 $N\,m\,kg^{-1}\,m^{-1}$. This was also slightly larger than that reported by most of the other studies, being 1.01 $N\,m\,kg^{-1}\,m^{-1}$ for subjects who were older than those of the present study (Gross et al., 1988; Pai et al., 1994; Schultz et al., 1992). The estimated peak ankle plantarflexor torque was 0.64 $N\,m\,kg^{-1}\,m^{-1}$, which is comparable with the 0.6 $N\,m\,kg^{-1}\,m^{-1}$ reported by Roebroeck et al. (1994) for younger individuals. The larger knee extension torque reported in this study could be related to the fact that our data also incorporated the measurements of chair-thigh contact forces.

When the normalized values in the present study were converted into absolute values, the hip peak flexion torque was 22 Nm, as compared with the 65 Nm reported by Fleckenstein et al. (1988) and 45 Nm by Pai and Rogers (1991) for younger individuals. The peak ankle dorsiflexion torque was 43 Nm in the present study, as compared with the 50 Nm reported by Fleckenstein et al. (1988) and Pai and Rogers (1991). Thus, the peak hip flexion torque was much smaller and the ankle dorsiflexion torque was slightly smaller than the previously reported values. This might reflect a tendency among young individuals to increase the hip flexion torque in order to enhance the horizontal momentum at the beginning of the movement. Older adults, however, might prefer a moderate initial peak hip flexion torque, while increasing the ankle dorsiflexion torque to control the increasing horizontal momentum and the CoM position during seat-off, a strategy that provides better postural stability during STS movement.

4.2. Mechanical and neurological deficits in PD subjects

PD subjects had a significant 40% reduction in the peak hip flexion torque when compared with the able-bodied subjects. This decrease in hip flexion torque could reflect an inability and/or inconsistency in recruiting motor unit activity (Dengler, 1986; Petajan, 1983), insufficient muscle force generated by the hip flexors (Berardelli et al., 1986; Frank et al., 2000; Teasdale et al., 1990) and/or increased co-activation of the antagonists (Beckley et al., 1991; Hayashi et al., 1988; Horak et al., 1992; Johnson et al., 1991). In addition, PD subjects demonstrated a slower build-up of the hip flexion and ankle dorsiflexion joint torques. Smaller hip flexion and ankle dorsiflexion joint torques could lead to difficulties in initiating STS movement, leading to a smaller horizontal velocity. It has been previously reported that PD subjects exhibited a smaller posterior excursion of the CoP during the initiation phase of STS (Mak and Evans, 1997). This implied a problem in anticipatory postural adjustment, leading to an inability to build up an adequate propulsive force to initiate the movement (Latash et al., 1995). Since STS is a self-perturbing task, the patients could choose to enhance postural stability during this task, by restraining hip flexion torques so as to limit the forward momentum. A smaller amount of hip flexion momentum can be advantageous in accurately bring the body CoM over the feet area during seat-off.

The fact that the measurements in this study were made on the left side only raises the question of bi-lateral symmetry. Previous studies have indeed demonstrated that elderly able-bodied subjects and patients with neurological or orthopedic disorders often demonstrate medio-lateral or bi-lateral asymmetry in joint torques and weight-bearing imbalance between the right and left sides (Levin et al., 1998; Isakov et al., 1992; Levin and Mizrahi, 1996). For instance, medio-lateral torque imbalance in normal subjects may originate from a minor orthopedic disorder, such as a length difference between the right and left legs. Muscle weakness on one side in PD subjects is expected to demonstrate similar effects. Yet, previous reports showed that differences between sides were found to be statistically insignificant in patients on medication (Steiger et al., 1996), and small during slow speed movements in moderately affected patients (Kakinuma et al., 1998).

The use of a similar initial posture in the protocol ruled out the possibility that the differences obtained between able-bodied and PD subjects originated from chair height or knee position. Similarly, with respect to the speed of ascent: the subjects were instructed to stand up at their natural speed. It is thus more likely that the differences in the kinematics and dynamics found between the two groups were due to differences in their behavioral paradigms. The latter revealed a reduced synchronized intersegmental (or interlimb) behavior in the PD subjects as compared to the control subjects, indicating a dysfunction at the level of global coordination (Serrien et al., 2000).

It is somewhat unexpected to find an insignificant difference between the groups in the hip and knee extension peak moments during STS. This could be due to the small number of subjects in this study. The PD subjects, however, took a longer time to reach the peak torques, which was consistent with the findings from upper-limb tasks of PD subjects, who demonstrated a deficiency in the rate of force production (Corcos et al., 1996; Godaux et al., 1992; Jordan et al., 1992; Kunesch et al., 1995; Phillips et al., 1994; Stelmach et al., 1989). In the present study, the slowness of STS in PD subjects could be due to the reduced hip flexion joint torque and the inability to generate the required joint torques as rapidly as the control subjects. These led to a decrease in acceleration and hence a decrease in the peak velocity of the CoM. In conclusion, the aforementioned findings suggest that a smaller hip flexion torque and a prolonged rate of torque production are significant mechanisms that contribute to the disruption of STS performance in PD subjects.

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Appendix A. Iterative estimation of joint torques using ground and seat reaction forces

A.1. Kinematics

A two-dimensional, four-joint, five-segment model, representing the feet, lower legs (shanks), upper legs (thighs), trunk and head was used for the analysis of the body dynamics in chair-rise as described in Fig. 1. The anthropometric dimensions in equations to follow (A.1)–(A.4), i.e., the components of the position vectors $\boldsymbol{p}_j$ and $\boldsymbol{d}_j$, and the inertia components of each segment (m_j, I_j), were evaluated in their respective local systems by scaling the subject height and mass with anthropometric coefficients taken from Winter (1990). All segments were assumed to be rigid, and to move only in the sagittal plane. According to the global frame orientation ($oxyz$), the z-axis is directed vertically downward; y is the horizontal axis in the anteroposterior direction; and the x-axis is perpendicular to the plane of motion (i.e. the medio-lateral direction), as described in Fig. 1. We assigned the generalized coordinates θ_i ($i = 1, \ldots, 4$) to represent the sagittal rotation of the segments about their local pivot axes x_i, with angular displacements measured anti-counterclockwise from the horizontal (y) axis. Definitions for the joint angles were as follows: θ_1—ankle, θ_2—knee, θ_3—hip and θ_4—neck (Fig. 1). Numerical differentiation of the time histories of the joint angles yielded angular velocities and accelerations. The Newton–Euler equations were subsequently used to convert the angular displacements and velocities into the linear velocity vectors of the segment's centers of gravity.

A.2. Dynamics

Linear and angular momentum equations were applied to each of the above segments to yield the system equations of motion. The linear momentum of the CoM of each segment, $\boldsymbol{L}_j$ reads

$$\boldsymbol{L}_j = m_j \boldsymbol{v}_j \tag{A.1}$$

where m_j and $\boldsymbol{v}_j$ ($j = 1, \ldots, 5$) are the mass and the linear velocity vectors (with two components) of the CoM of the jth segment, respectively. In a similar manner, the angular momentum (in the y–z plane) of the segment about its CoM follows:

$$H_j = I_j \omega_j \tag{A.2}$$

where I_j is the inertia component of the jth segment relative to its CoM and ω_j designates the angular velocity component of the segment (a scalar). Next, the Newton–Euler equations of motion were applied to each of the segments to compute the forces and moments in the joints:

$$\frac{\mathrm{d}\boldsymbol{L}_j}{\mathrm{d}t} = \boldsymbol{F}_{j-1} - \boldsymbol{F}_j - m_j \boldsymbol{g} \tag{A.3}$$

$$\frac{\mathrm{d}H_j}{\mathrm{d}t} = \|\boldsymbol{d}_j \times \boldsymbol{F}_{j-1}\| - \|\boldsymbol{p}_j \times \boldsymbol{F}_j\| + \tau_{j-1} - \tau_j \tag{A.4}$$

The left-hand part in Eq. (A.3) represents the time rate change of the linear momentum of the jth segment CoM, and the left-hand part of Eq. (A.4) represents the time rate change of the angular momentum of the segment about its CoM. $\boldsymbol{F}_j$ and $\boldsymbol{F}_{j-1}$ are the intersegmental forces (with two components) acting at the top-most (proximal) and the bottom-most (distal) ends (joints) of the jth segment, respectively. Similarly, τ_j and τ_{j-1} are the joint torques (scalars) acting on the same respective ends. $\boldsymbol{p}_j$ and $\boldsymbol{d}_j$ are the position vectors (with two components) connecting the CoM of segment j to its proximal and distal joints, respectively.

A.3. The iterative algorithm

Eqs. (A.3) and (A.4) yield a set of 15 ($= 3 \times 5$) scalar equations. There are potentially 18 unknowns: 12 ($= 3 \times 4$) force components and torques acting at each of the joints; and 6 ($= 3 \times 2$) unknowns at the body contact with the supports. However, since the chair-thigh and foot-ground reaction force and point of application were actually measured, the system was over-determined, allowing us to use the force plate measurements under the feet for comparison with the model results. It should be reminded, though, that differences between force-plate measurements and model estimated results may generally arise from errors in positioning the instantaneous centers of rotation of the ankle and knee, estimating the anthropometric properties of the segments, deformability of the trunk during movement, synchronization errors between force-plate and motion analysis system and digitization errors of the joint positions. Quantitative evaluation of the difference between the model estimated and the measured components of the force and CoP (platform A) was obtained by using the sum of the square errors (SSE):

$$\mathrm{SSE} = \sum_{k=1}^{N} (q_{\mathrm{mod}el}[k] - q_{\mathrm{meas}}[k])^2 \qquad \text{(A.5)}$$

where $q_{\mathrm{model}}[k]$ and $q_{\mathrm{meas}}[k]$ are the model and measured values at time point k, respectively, and N is the number of data points in the time history of the record. The iteration process allowed for the readjustment of the positioning of the centers of the ankle and the knee from their predetermined initial positioning to minimize SSE. In the iterative process, the following constraints were imposed: (i) the re-estimated (ith iteration) location of the ankle along the midline axis of the foot should not exceed the anatomical boundaries of the ankle joint; (ii) the model-predicted AP trajectory of the CoP should remain within the boundaries of the feet; and (iii) the re-estimated location of the thigh-seat reaction force, F_1 should remain within the boundaries of the seat. The first requirement was that the positioning errors along the midline axis of the foot of the intercept point between the talocrural and midline foot axes should be less than $0.12 l_{\mathrm{foot}}$ (Levin et al., 1998), where l_{foot} denotes the distance from the tip of the big toe to the posterior edge of the calcaneus. Within these values, the tibial talus and calcaneus were found to overlap (Inman, 1976). The two remaining constraints were indicated for each subject prior to the onset of data acquisition and were kept similar for each trial.

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