SOME BIOMECHANICAL CONSEQUENCES OF VARYING FOOT PLACEMENT IN SIT-TO-STAND IN YOUNG WOMEN

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ABSTRACT. The position in which the feet are placed prior to the start of sit-to-stand affects the distance to be moved by the body mass forward over the feet. The purpose of this study was to examine the effect of three foot placements, back, preferred and forward, on biomechanical features of the action. Six subjects were videotaped standing up from a seat with feet on a forceplate. XY coordinates and forceplate data were synchronised and kinematic and kinetic variables analysed. A forward foot placement significantly affected both the pre-extension and extension phases of the action. Under this condition, it was evident that body mass was moved the greater distance horizontally: *i*) by increasing the speed and amplitude of trunk flexion; and *ii*) by increasing the time over which the shank rotated forward at the ankle. Vertical movement of body mass was achieved largely by a 50% increase in peak moment of force at the hip. It is concluded that a forward foot placement would adversely affect the ease of standing up for individuals with leg muscle weakness and poor motor control. Since the altered biomechanical characteristics may place additional strain on the hip, patients should be cautioned to avoid standing up with feet forward after hip replacement surgery.

Key words: biomechanics, kinematics, kinetics, sit-to-stand, foot placement, support moment, joint moments, rehabilitation.

This study was designed to examine the effects of foot placement on certain biomechanical characteristics of sit-to-stand (STS) in young healthy adults. Standing up from the seated position is a common activity of daily living. It comes naturally and easily to able-bodied healthy individuals. However, elderly subjects have been reported to have difficulty with the action (3, 15, 16). In addition, standing up independently and efficiently has been reported to be a major problem for people with motor control disability and weakness of the lower limbs, and a foot placement which is not optimal may make the action impossible (8, 10).

In STS, the feet make up the fixed segments over which other body segments pivot as the body mass moves from over one base of support to another. Foot placement, therefore, specifies the distance forward over which the body mass must be moved as it shifts from over the thighs to the feet and must consequently be a major factor influencing the timing and pattern of force production in the lower limbs. Knowledge of the effects on the biomechanics of an action when it is performed under different conditions in able-bodied individuals enables the development of effective training strategies for rehabilitation of the movement-disabled individual (8, 10).

The ability to stand up, since it is critical as a prerequisite for walking, is essential to independence in daily life. Previous studies of STS have investigated the effects of various conditions, such as chair height (22), chair design (4, 5, 28), initial leg position and head posture (12, 26), extent of trunk movement (25), arm use (7) and speed (19).

In most studies of STS, foot position has either been standardised or subjects have been able to place their feet wherever they preferred. Studies in this latter group have found that subjects tend to place their feet back so that the foot lies posterior to the knee (27, 28). In one study in which elderly subjects stood up from a chair specially designed to aid rising, it was noted that the subjects had greater difficulty standing up from the special chair than from a standard one. One feature of the special chair was the small amount of clearance under the front of the seat which prevented foot placement backward (28). Another study reported that elderly and arthritic subjects complained that they did not have enough room to place their feet under their chair. A forward foot placement has been reported in one study (12) to affect the moments of force at the hip.

In the present study, certain kinematic characteristics of trunk, thigh and shank segments were examined as subjects stood up under three different foot placement conditions. Many previous studies of STS have investigated the effect of an experimental manipulation on one or two lower limb joints (12, 14). Although the understanding of individual joint forces is crucial, for example, in the design of joint prostheses, physiotherapists training individuals with movement disorders to stand up need also to understand how the lower limb works as a coordinated unit to perform the action; that is, how the trunk and the three segments of the lower limb cooperate to perform the action of raising the body from sitting to standing. Therefore, in addition to examining individual joint moments of force, support moment of force (29), an overall measure of force production which enables the lower limb to be examined as a single unit (25, 29), was also investigated.

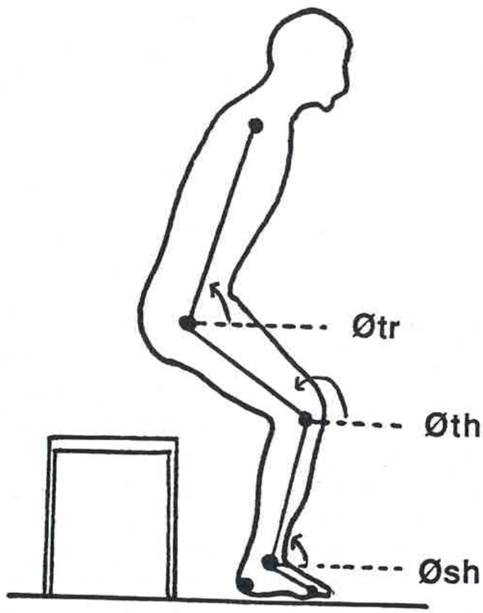

Fig. 1. The four-segment model used in the kinematic analysis showing absolute angles. All segments are defined as positive in a counter-clockwise direction. Øtr = trunk (hip); Øth = thigh (knee); Øsh = shank (ankle).

METHODS

Subjects and procedure

Six female subjects aged between 18 and 25 years (M = 21.3 years, SD = 1.03) and of average height (M = 1.67 m, SD = 0.05) and weight (M = 56.75 kg, SD = 9.11) volunteered to take part in the study. They were informed about the nature and purpose of the study and gave their written consent. All procedures were in accordance with approved ethical guidelines.

The methodology used has been developed in our laboratory and reported in detail elsewhere (7, 25). Subjects sat on a height-adjustable seat with both feet on a Kistler forceplate (type 9281, 400 × 600 mm). They sat with their ischial tuberosities on a pressure-sensitive switch, with a sensitivity of 1 N, attached to a light. The signal from the switch recorded loss of contact with the seat (called thighs-off) and was recorded with forceplate data on a computer and on the videotape. Reflective markers were placed over the lateral gleno-humeral joint, greater trochanter, head of fibula, lateral malleolus, heel, head of fifth metatarsal. These defined a four-segment model composed of head-neck-trunk (called trunk) segment, and thigh, shank and foot segments (Fig. 1). Subjects were videotaped by a National WVO-F10 videocamera with fixed sampling rate of 25 Hz and shutter speed of 1/1000 s from the right side. Forceplate data were sampled at 200 Hz and input to a computer, giving 1000 samples of forceplate data per channel.

Subjects stood up from a standardised starting position: trunk erect and hands in lap. They were instructed to look straight ahead at a target placed at standing eye height to standardise head movement. Subjects were also instructed to stand up at a comfortable speed with weight distributed evenly on both feet. By filming from one side with both feet on one forceplate the assumption was made that the action is performed symmetrically. This assumption is considered reasonable for young healthy adults and is commonly made in studies of STS (12, 16, 20, 27). This assumption has lately been questioned by Lundin and colleagues (16) and it is clear that there may be asymmetry between the movements at individual lower limb joints (20). For the purpose of this study, the two lower limbs were modelled as a single functional unit.

Seat height was standardised to 110% of each subject's lower leg length. Foot placement was standardised for each condition (Fig. 2): feet back (FB) – heels aligned with front legs of seat; feet forward (FF) – shank perpendicular; and feet preferred (FP). For this latter condition, foot placement spontaneously selected over three practice trials was averaged for each subject. Order of conditions was counterbalanced and subjects had four practice trials under each condition before data collection commenced.

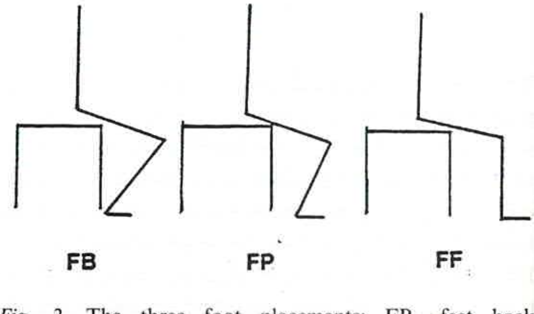

Fig. 2. The three foot placements: FB = feet back, FP = preferred position, FF = feet forward.

Data analysis

X and Y coordinates were digitised manually at a sampling frequency of 25 Hz using a system which included a digitising pad and videomixer.

The videomixer enabled two video signals, one from the digitising pad showing the cursor, the other from the VCR showing the stop-frame image of the subject, to be superimposed on the monitor (1, 26). The video and forceplate data were synchronised using a LED timing device connected to a computer and in view of the camera. During the digitising process, the timing box number was recorded for each frame and subsequently transferred to computer files so that kinematic analysis was matched frame by frame to the kinetic analysis (26).

Segmental kinematics and kinetics were determined from the coordinate and forceplate data. Three segment angles (trunk, thigh and shank, called hip, knee and ankle for ease of description) were calculated as absolute angles in space in a four-segment three-joint system. Basic linked-segment equations, utilising kinematics, anthropometrics and ground reaction forces (31) gave the joint moments of force at the three joints. Joint moments were reported following the right-hand rule, with hip and ankle moments extensor when −ve and knee moments extensor when +ve. Support moment of force (Ms) was calculated as an algebraic summation of moments at knee, hip and ankle, $Ms = Ma + Mk + Mh$, where Ma, Mk and Mh are the moments at ankle, knee and hip expressed in N·m (29). Kinetic data were normalised to each subject's body mass by dividing the moment of force by each subject's body mass. In order to allow for inter-trial and inter-subject comparison, all output data were prepared on a normalised time base, as a percentage of the extension phase (with thighs-off as zero). For descriptive purposes, repeat trials, with amplitude normalised by dividing by each subject's body mass (29), were ensemble-averaged and the variability expressed in the form of a coefficient of variation (CV) (30).

Ensemble averaging involves calculating the average and standard deviation of two or more curves at each 2% of the action. The CV of the ensemble-averaged data provides the root mean square variance of the curve relative to the curve's mean amplitude expressed as a percentage (26, 30).

Three events were used to provide reference points for kinematic and kinetic data: movement onset was defined as the moment at which horizontal velocity of the shoulder marker equalled or was greater than 0.1 m/s; thighs-off as loss of contact with the seat switch; movement end as the moment when horizontal velocity of the hip marker was equal to or less than 0.1 m/s. These events enabled the action to be divided into two phases: a pre-extension phase (movement onset to thighs-off) and an extension phase (thighs-off to movement end).

The variables of interest were examined across conditions using planned contrasts within repeated measures analyses of variance, with Scheffe's F-test used for post-hoc comparisons. The significance level was set at $p < 0.01$.

RESULTS

Movement time

As the feet were placed further forward, the overall time taken to perform the action increased. This increase was principally due to an increase in the extension phase, with the pre-extension phase taking a similar time at each foot placement (Fig. 3). Post hoc analysis indicated that, for overall movement time, there was a significant difference between both FB and FP and FF, and, for the extension phase, between FB and FF.

Pre-extension phase

Angular kinematics. Although there was no significant difference between conditions in the duration of the pre-extension phase, there were significant changes in the amplitude of the two pre-extension phase kinematic variables examined. Hip flexion displacement and velocity both increased significantly as feet were placed further forward (Fig. 4). Repeated measures ANOVA found significant main effects of displacement and velocity and post-hoc analysis indicated that there were significant differences between each foot placement for both displacement and velocity.

Extension phase

Joint displacements and moments of force. The pattern of individual joint moments showed a similarity across conditions in that hip moment was extensor throughout the extension phase with the moment at the knee changing from extensor to flexor in the second half of the extensor phase (Fig. 5). The peak values of extensor moments at the three joints, however, changed according to condition (Table I). A repeated measures ANOVA revealed a significant difference between hip and knee/ankle complex and between the knee and ankle. Fig. 6 shows the significant joint by condition interaction, which indicates that peak hip moment increased whereas ankle and knee moments decreased as the feet were placed further forward. The timing of peak moments was also marked by differences in FF compared to the two other conditions. In FB and FP, the peak moments were produced at or around thighs-off. In FF, the peak values occurred in the sequence hip, knee, ankle. It is evident, therefore, that when the feet were forward, vertical propulsion of the body mass at thighs-off was largely due to the moment at the hip, with negligible moments at the knee and ankle.

The changes which occurred in joint moments across conditions were reflected in the changes in joint angular displacement due to the different foot positions (Fig. 7). Although extension took place at

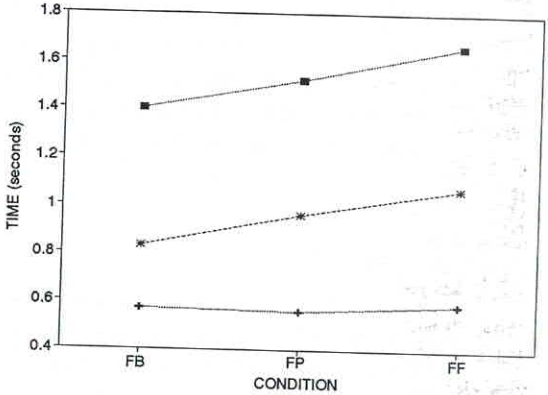

Fig. 3. Mean total movement time (■), duration of pre-extension (+) and extension (∗) phases in the three conditions.

the hip and knee throughout the extension phase in all three conditions, the extent and timing of motion at the ankle joint appeared to be affected by foot position. When the feet were forward, the amplitude of active ankle dorsiflexion was greater than when the feet were further back and the ankle dorsiflexed throughout the first 34% of the extension phase. As a consequence, in this condition, the shank segment would have contributed little to the production of the extensor force required to propel the body mass vertically at thighs-off. As is shown in Fig. 5, peak moment at the ankle at thighs-off in FF was on average less than 0.2 N·m/kg.

Support moment of force (SM) was examined as a means of determining the overall force produced through the lower limbs in the extension phase irrespective of the variability produced at the individual joints. Peak SM occurred at or just after thighs-off in each condition (Fig. 8), the peak value decreasing the feet were moved further forward (Table Although there was a significant main effect of co dition, post-hoc analysis indicated that the differen lay between the two most extreme foot placemen FB and FF.

Interestingly, when individual subject and trial da were examined, there was some variability betwe subjects in the pattern of SM when the feet we placed forward. Three distinct patterns could identified (Fig. 9). Pattern A, with a distinct pea occurring at thighs-off, was similar to that in t graphs for the FB condition and occurred in all fo trials of one subject. Pattern B, similar to that in t FP condition, where the peak value occurred 5% in the extension phase, was the most common and wa evident in all trials of three subjects. Pattern C, distinctly different pattern with the occurrence of double peak, occurred in all trials of one subject. Th only subject who demonstrated more than one patter had two trials each of pattern B and C. Visua inspection of the videotapes showed that in the trial which illustrated pattern C, subjects appeared to lif off the seat then pause before extending into th standing position. The two peaks can be taken t illustrate the initial lift-off and the added burst o extensor thrust which occurred a little later.

DISCUSSION

The results of this study provide evidence of the way in which the biomechanical characteristics of STS can be influenced by the initial foot placement. It is intuitively obvious that foot placement and the

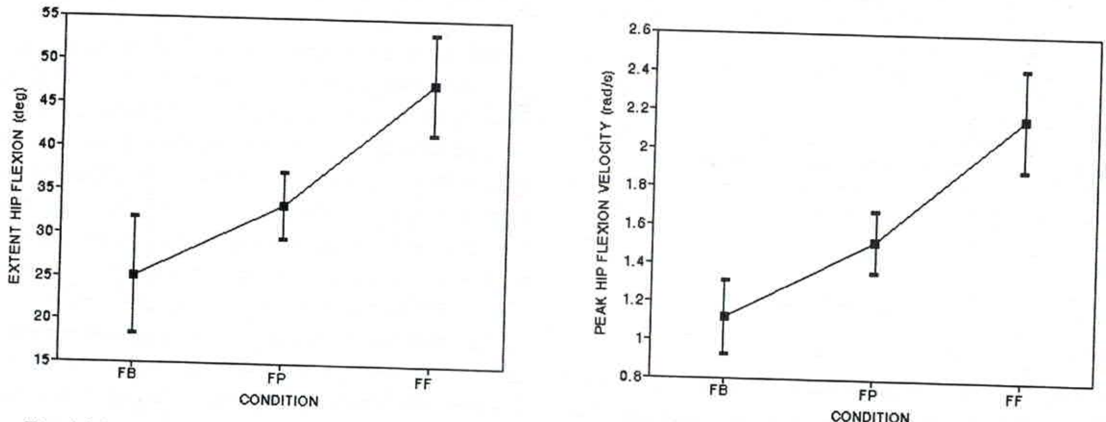

Fig. 4. Mean peak hip flexion displacement (left) and mean peak hip flexion velocity (right) in the three conditions.

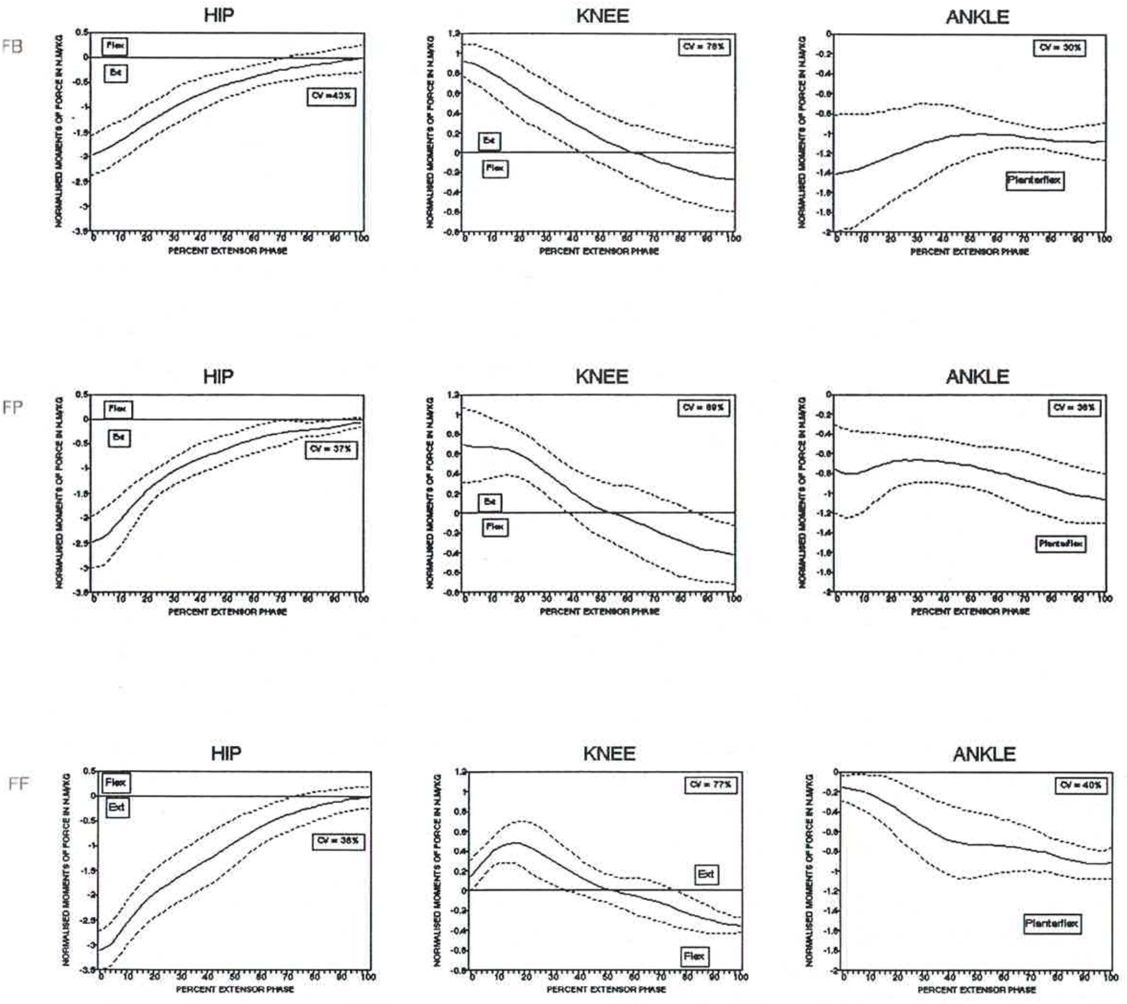

Fig. 5. Ensemble-averaged profiles of normalised joint moments of force for four trials on six subjects under three foot placement conditions. 0% = thighs-off; 100% = movement end. Coefficients of variation (CV) are mean standard deviations (*dotted line*) as a percent of the mean (*solid line*). Hip and ankle extensor moments are indicated by negative values, knee extensor moment by positive.

dynamic characteristics of STS are affected by factors such as the environment in which the action takes place (e.g., seat height, chair design) and the task being performed (e.g., standing up to walk out the door or to put a glass on a table). Individuals in daily life must be able to adapt to all these challenges. However, the further forward the feet are placed in sitting, the more the distance to be moved by the body mass from over the seated base of support to the feet increases. If the feet are too far forward, it may no longer be possible to stand up.

It appears from the present study of three different foot placements that the mechanism enabling the body mass to move forward the increased distance

Table I. *Means and standard deviations for peak extensor moments of force under the three conditions*

FB: feet back, FP: feet preferred, FF: feet forward.

Peak moments of force (N·m/kg)	Condition			
		FB	FP	FF
Peak hip moment	M	2.0	2.5	3.2
	SD	0.2	0.1	0.2
Peak knee moment	M	0.9	0.8	0.5
	SD	0.1	0.1	0.1
Peak ankle moment	M	1.6	1.1	1.0
	SD	0.1	0.2	0.1
Peak support moment	M	4.3	4.0	3.6
	SD	0.8	0.7	0.4

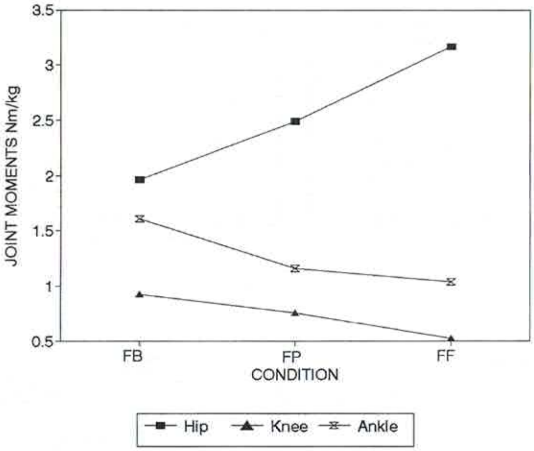

Fig. 6. Mean peak normalised moments of force at hip, knee and ankle under the three conditions.

included an increase in the amplitude of both hip flexion and velocity in the pre-extension phase. In addition, when the feet were forward, the movement of the body mass forward by the action of dorsiflexor muscles acting on the shank would be critical to moving the body mass forward. Indeed this study shows that ankle dorsiflexion continued through the first one-third of the extension phase under this condition.

It has been suggested that angular momentum generated by the forward rotation of the trunk segment is harnessed into the vertical projection of the body mass in the extension phase (24). This transfer in momentum was postulated to decrease the amount of lower extremity muscle force required at thighs-off. Given the need to shift the body mass a considerable distance forward when the feet were placed forward, the higher peak hip flexion velocity and the lower peak support moment found in the present study could be taken to support the suggestion of Schenkman and colleagues (24) since SM represents a summation of extensor forces produced through the limb. However, the difference in the timing of individual peak joint moments of force across the three conditions suggests that another factor may also have resulted in producing the lower mean peak value of support moment when the feet were placed forward. Although when the feet were back, moments of force at hip, knee and ankle peaked together around thighs-off, when the feet were forward, there was a marked difference in the distribution of joint moments across the limb at thighs-off. Individual joint moments peaked sequentially, peak hip moment at thighs-off, knee moment at 40% and ankle moment toward the end of the extension phase. As such, the lower peak support moment (a summation of the three joint forces) in the FF condition may merely reflect this sequencing of extensor forces.

As well as affecting the distribution of moments at the start of the propulsive extension phase, the amplitude of joint moments also varied significantly according to foot placement. On average, hip extensor moment of force was greatest when the feet were forward, a finding reported also by Fleckenstein and colleagues (12) in a study of the effects of varying initial knee angle, and therefore, relative foot position. In the present study, at thighs-off, when the major lower limb propulsive thrust into extension is necessary, force produced at the hip appeared to be the major contributor to the vertical movement of the body mass when the feet were forward. Knee and ankle moments at this time were negligible. Considering linked-segment dynamics, both trunk and thigh segments would have contributed to the moment produced at the hip via both monoarticular and biarticular muscles. It is only possible to speculate on the muscle contribution to the large hip moment and small knee moment at thighs-off. However, it is likely that the action of the monoarticular vasti muscles was at this time directed at extending the hip by rotating the thigh segment in a clockwise direction. This action, together with the action of hip extensor muscles (the monoarticular gluteus maximus and biarticular hamstrings), acting to rotate the trunk in a counter-clockwise direction, would have resulted in the large hip extensor moment. That is to say, hip extension may have been brought about by the action of muscles that do not span the joint as well as by those that do (32) illustrating the complexity of multisegmental movement.

The large hip moment when the feet were forward would have been a critical factor in the prevention of limb collapse as well as in the required vertical propulsion of the body mass, given the mechanical disadvantage at the knee and ankle. However, the large hip moment may also have contained a significant braking component. The increased hip flexion displacement and velocity required to move the body mass forward over the increased distance between seat and feet would have generated an increase in horizontal momentum of the body mass with considerable threat

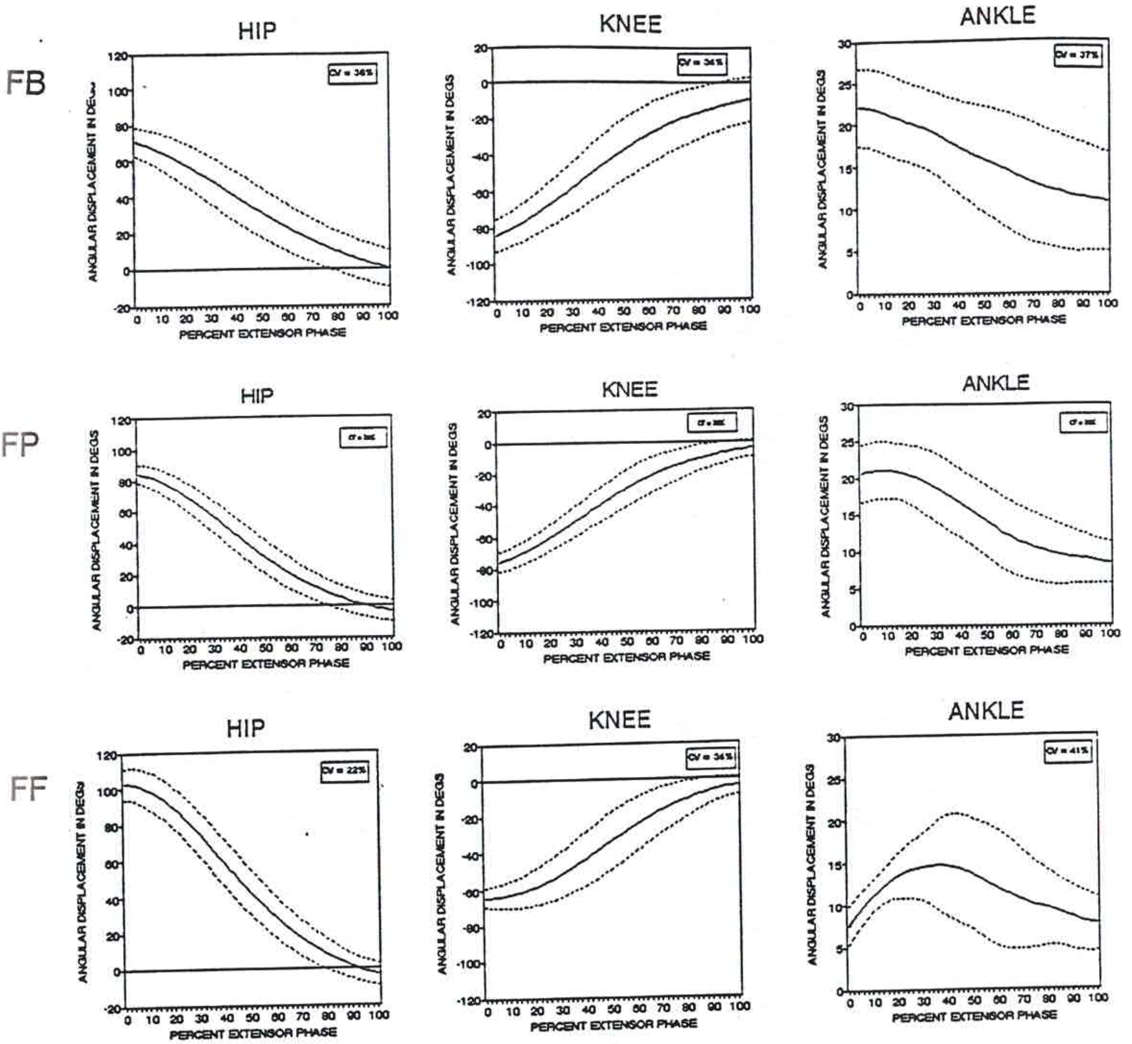

Fig. 7. Ensemble-averaged profiles of angular displacements of the trunk (hip), thigh (knee) and shank (ankle) segments for 36 trials under the three foot placement conditions. Coefficients of variation (CV) are mean standard deviations (*dotted line*) as a percent of the mean (*solid line*). 0% = thighs-off; 100% = movement end.

to stability. The large hip extensor moment may therefore reflect a braking effect on the horizontal movement of the body mass. Pai and colleagues (21) have recently shown that horizontal momentum is typically more constrained than vertical momentum in STS, providing, they suggest, a mechanical constraint on the control of balance. Momentum in the horizontal direction governs the fate of the projection of the body mass relative to the base of support, increased horizontal momentum requiring additional muscle force to provide adequate braking.

The continuing ankle dorsiflexion into the extension phase when the feet were forward would have prevented the plantarflexor (extensor) muscles from contributing to vertical propulsion at thighs-off since the dorsiflexors were rotating the shank segment in a clockwise direction. This is in contrast to the two relatively posterior placements, when the anterior tibial muscles would have been isometrically active (23). The findings also suggest that extensor force production by the shank segment was delayed until the body mass was sufficiently forward, thus ensuring that subjects would not over-balance backward. It is evident, therefore, that, at least when the feet are placed forward, the lower leg muscles are responsible not only for the control of balance (11, 23), but are

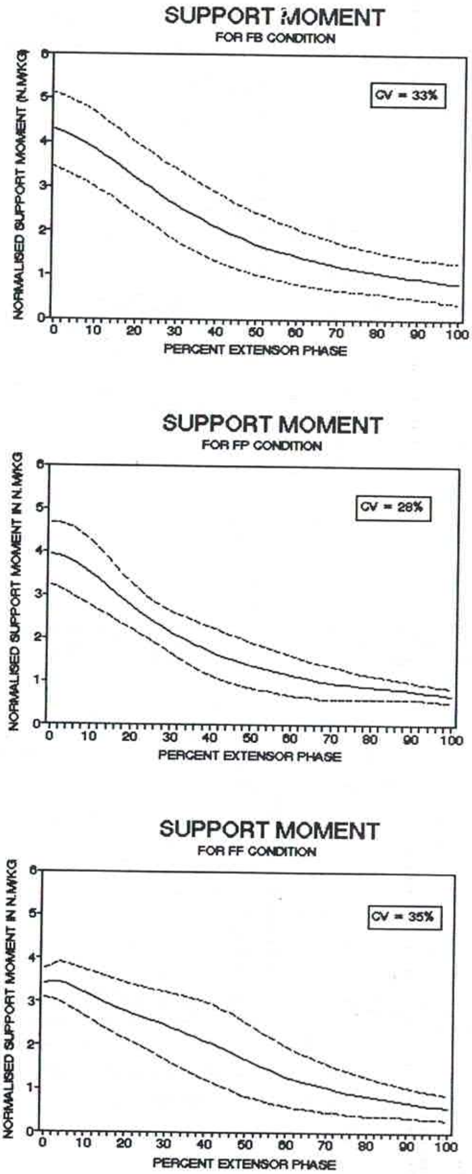

Fig. 8. Ensemble-averages of normalized support moment of force for four trials on six subjects under the three conditions.

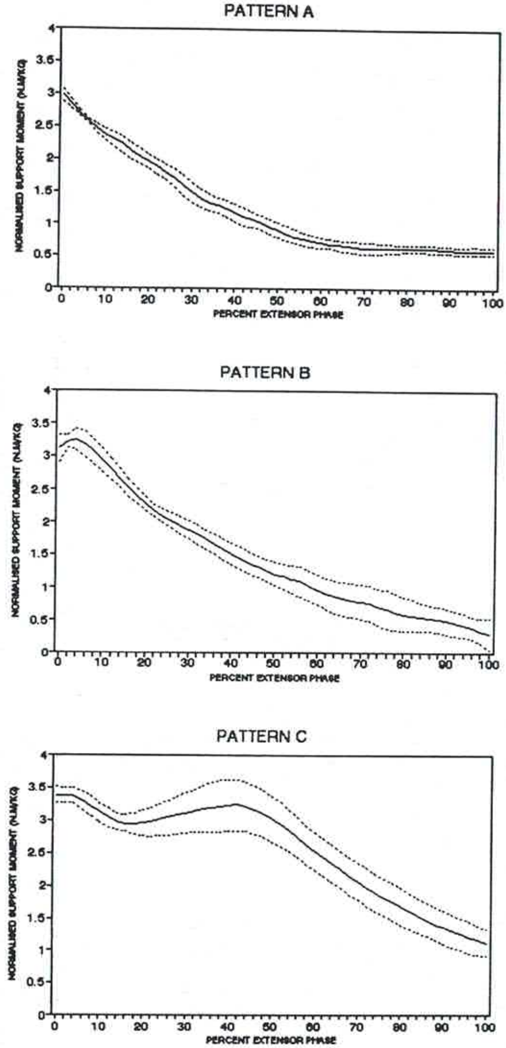

Fig. 9. Ensemble-averages of three patterns of normalised support moment in the FF condition. A: $n = 4$; B: $n = 14$; C: $n = 6$ trials.

also involved initially in the transport forward and subsequently in the vertical propulsion of the body mass.

When the feet were forward, the duration of the extension phase was significantly longer than when the feet were back. Furthermore, the variation between subjects in the pattern of support moment when the feet were forward suggests that subjects adapted to this more difficult condition by using different strategies. Analysis of the pattern of support moment used by two subjects (pattern C) shows that a relatively high value of support moment was sustained over a large proportion of the extension phase in order to bring the body into standing. For these two subjects at least, this raises the question of the efficiency of standing up with the feet forward. It could be assumed that for a propulsive action such as STS, a short strong burst of muscle effort at the start of vertical propulsion (thighs-off) may be more energy conserving than the more uniform distribution of effort evidenced in pattern C. Pai and colleagues (20) have pointed out the dynamic necessity for the generation of a propulsive impulse to generate the necessary movement vertically of the body mass.

Mean peak support moment in the feet back condition (4.3 N·m/kg) was a little lower than the mean values (4.7 N·m/kg) reported in two previous studies of young male subjects with familiar foot placements and using a similar methodology (7, 25). Since the movement duration was similar in the three studies, the difference may be associated with gender-based strength differences as it has been suggested that strength differences would affect the pattern of movement used in sit-to-stand (28).

The findings from this study have implications for future research into STS as well as clinical implications. Since foot placement affects certain biomechanical characteristics of the action, foot position as well as seat height, which varies the relationship between thigh and shank segments, need to be standardised and reported for results to be compared across investigations. If a preferred foot placement is considered to be desirable, this position can be standardised, as in this study, as a means of reducing intra-subject variability. Interestingly, the preferred placements in this group of subjects were on average within the range of 8 to 16 cm back from a perpendicular line drawn from the knee to the floor with the subjects seated.

In terms of clinical intervention, the results support the view that standing up with the feet relatively forward would not allow for optimum performance in certain disabled populations (10). For example, the relatively high peak hip flexion velocity attained in the pre-extension phase would be difficult to achieve in many patients with lower limb motor dyscontrol and muscle weakness, and has been shown to be so following stroke (2). Given the association between ability to stand up and strength of hip extensor muscles reported in a group of elderly subjects (14), it could be expected that this group may also have difficulty generating sufficiently high hip extensor forces. The need for high peak hip extensor moments and the increased amplitude and velocity of hip flexion with the feet forward would be contraindicated for people following hip replacement surgery. Rising from a seat has been reported to produce three times the pressure at the hip joint as that produced during walking (13) and this value may be greater when the feet are forward. It is also considered that excessive hip flexion may loosen the prosthesis (6).

There are two major clinical implications from the present study. The foot position used by a disabled person during motor training in the initial stages of rehabilitation should be relatively posterior (approximately 10 to 12 cm posterior to a perpendicular line drawn from the knee when sitting on a standard seat) in order to facilitate performance of the action, particularly when leg extensor muscles are weak (10). For the feet to be placed posteriorly requires that range of motion at both knee and ankle be sufficient to allow for the necessary flexion at these two joints. Given that chairs (including wheelchairs) designed for the elderly as well as for the disabled typically restrict backward placement of the feet, a second implication is that consideration should be given to ease of rising to standing as well as to sitting comfort in the design of chairs for both disabled and elderly populations.

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