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RISING FROM A CHAIR BY A SPRING-LOADED FLAP SEAT: A BIOMECHANICAL ANALYSIS

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ABSTRACT. Nine healthy male subjects and 8 patients with unilateral knee osteoarthrosis were studied while rising from a chair with and without the aid of a spring-loaded flap seat. The seat force started at 216 N and decreased with increasing seat angle. Ground reaction forces and motion were recorded using a force plate and video. EMG of the left vastus lateralis muscle was recorded. The effort when rising was estimated by the Borg scale. The use of the flap seat reduced the mean peak knee moment for the healthy subjects from 73 to 41 Nm and for the patients from 55 to 33 Nm. The decreases were significant at a $p < 0.001$ level. The patients also reduced their hip moment significantly from 50 to 35 Nm ($p < 0.005$). The muscle activity in the vastus lateralis was also significantly lower when the flap seat was used ($p < 0.005$). When the theoretically maximal friction force between seat and subject was added, the effective seat force decreased more slowly, and the relation between its horizontal and vertical force components was more advantageous. The patients estimated greater effort reduction than the healthy subjects comparing the ordinary chair and the flap seat. Conclusion: a spring-loaded flap seat can reduce knee and hip load and can thus be useful for people with knee osteoarthrosis.

Key words: rising, flap seat, joint load, knee osteoarthrosis, EMG, biomechanics.

Osteoarthrosis of the hip and knee joints is a common and often painful disease (1, 14, 18, 19, 20). The pain usually increases with increasing joint load, and the joint affected benefits from load reduction as this reduces pain and may prevent further joint destruction (1, 2, 15). People with such disabilities often have difficulties in rising from chairs (5, 10), and often there is accompanying muscle weakness that increases with age.

Different types of rising aids have been designed (e.g. 4, 8), and in this study we analyze one of the models. It is a spring-loaded "flap" seat which can be attached directly to the seat of an ordinary chair. The aim of the study was to investigate how the use of such a seat during rising affects the hip and knee joint loads and the activity in one of the big thigh muscles, and to analyze the forces exerted by the seat. At first we used young healthy subjects to obtain a reference material and to see that the method worked. Then we involved potential users, patients with knee osteoarthrosis, in the study. The healthy subjects were also used to establish the mechanical properties of the seat.

MATERIAL AND METHODS

Subjects

Nine healthy subjects, all men aged between 20 and 32 years, participated in the study. Their average height and weight were 1.80 m (range 1.73–1.94 m) and 73 kg (range 64–87 kg) respectively. The subjects (medical students) were all without functional disabilities of the lower limbs.

Patients

Eight patients, 4 men and 4 women, with moderate osteoarthrosis of the medial compartment of the knee, scheduled for surgical treatment, constituted the patient group. Their mean age was 70 years (range 65–79 years), mean height 1.71 m (range 1.58–180 m) and mean body weight 73 kg (range 50–98 kg). The degree of osteoarthrosis was assessed from radiographs of the knee joint in anteroposterior weight bearing projection, and in lateral projection and the knee joint in semiflexion. Two patients had bilateral knee osteoarthrosis and one of these patients also had slight pain from her hips. In 4 patients the right knee was scheduled for surgery, and in 4 patients the left knee. Each patient was physically examined to establish a Hospital for Special Surgery (HSS) knee score, with a total of 100 points (Gore et al. (9)). The mean HSS score was 60 points (SD 13 points) which is classified as a poor knee function. Knee pain during walking and at rest was assessed using a 10-cm long Visual Analog Scale (VAS), with 0 cm as no pain and 10 cm the worst possible pain. The mean knee pain during walking was graded as 6 cm (SD 2 cm) and at rest as 3 cm (SD 3 cm) by the patients. On physical examination all knees were stable. All patients could flex their knee to 100° or more and no patient had a flexion contracture of more than 10°.

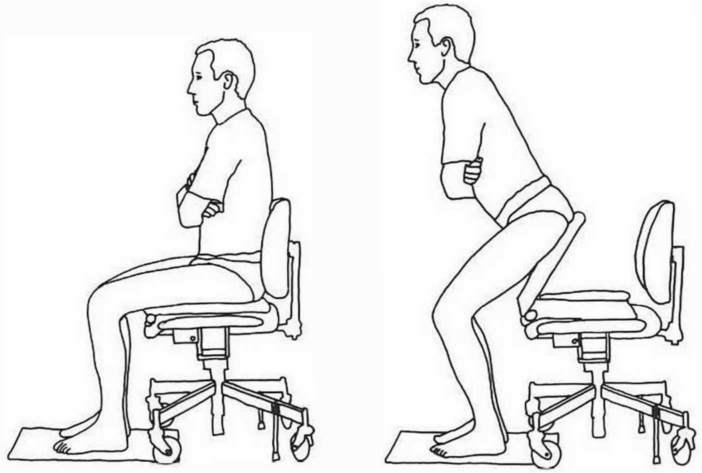


Fig. 1. Subject rising with the flap seat.

Flap seat and seat forces

The flap seat studied was of a type which did not allow adjustment of the force. It consisted of two wooden flaps (40×40 cm) connected at the front edge by hinges and two strong springs. The lower flap was attached horizontally to the seat of a chair, and the upper one formed the seat on which the subject sat. When sat upon, the flaps were held together, and during rising spring forces on the upper flap helped to push the subject upward/forward. The upper flap was padded and covered with cloth that increased the friction and prevented the subjects from slipping. The seat flap was stopped by a strap at 70° from the horizontal. Fig. 1 shows one of the subjects rising with the flap seat.

The flap seat was attached to the seat of a chair and the height was adjusted to knee level (lateral epicondyle). Subjects and patients practised rising until they felt they could use the device in an optimal way. For comparison, rising was also performed from the same height without using the flap seat. The rising speed was not strictly controlled but the participants were told to try to rise at the same speed with and without the flap seat. Arm rests were not used but the patients were allowed to hold their hands on low arm rests when using the flap seat in order to maintain their balance.

To establish the mechanical properties of the flap seat, the exerted spring force was measured with an electric dynamometer connected to the back edge of the seat flap, from the horizontal position and every 10°. The seat-to-trunk force was assumed to act through the subjects' ischial tuberosities at a calculated mean distance of 25 cm from the front edge of the flap. The seat-to-trunk force at this point was calculated according to the moment equation $F \times 0.40 = F_1 \times 0.25$, where F is the force at the back edge of the flap at different seat angles and F_1 the corresponding force through the ischial tuberosities. The position of the ischial tuberosities on the seat was constant during the movement.

Besides the seat normal force created by the springs there is also a friction force between the subject's behind and the flap. When this friction force is added to the seat normal force, a new force, the effective seat force, is created. This force is larger and more vertical than the seat normal force. Fig. 2*a* shows the seat normal force N, the maximal friction force F, and the effective seat force P, at a flap angle of 30°. The effective seat force is created when the theoretically maximal friction force for each flap angle is used. We have only analyzed the case when the friction force is directed upward and will assist in rising. Thus, we have assumed that the forces from the seat are utilized in a theoretically optimal way. The

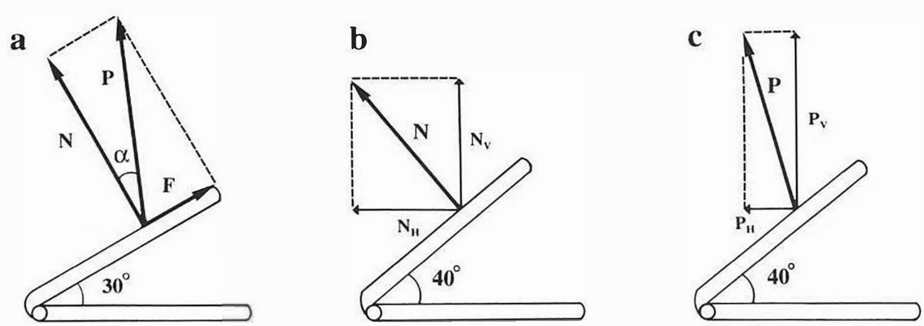


Fig. 2. (*a*) Forces between seat and subject. N is the seat normal force exerted by the two springs, F the theoretically maximal friction force and P the effective seat force which is the sum vector of N and F. α is the angle between N and F. (*b*) The seat normal force N, resolved into vertical N_V and horizontal N_H components. (*c*) The effective seat force P, resolved into vertical P_V and horizontal P_H components.

friction depends on the friction coefficient for the two surfaces. To calculate this, two of the subjects were pulled over the horizontal flap and the force needed (friction force) was measured. The friction coefficient could then be calculated according to the formula q = F/N (q = friction coefficient, F = friction force, N = force of gravity between subject and flap).

The angle α between the seat normal force N and the effective seat force P (Fig. 2*a*) was calculated using the following equations:

$$F = \sin\alpha \times P \tag{1}$$

$$N = \cos\alpha \times P \tag{2}$$

$$q = (\sin\alpha \times P)/(\cos\alpha \times P) = \sin\alpha/\cos\alpha = \tan\alpha \tag{3}$$

$$\alpha = \arctan q \tag{4}$$

When the angle α was known, the effective seat force P was calculated using the following equations:

$$\cos\alpha = N/P \tag{5}$$

$$P = N/\cos\alpha \tag{6}$$

Force recording and moment calculation

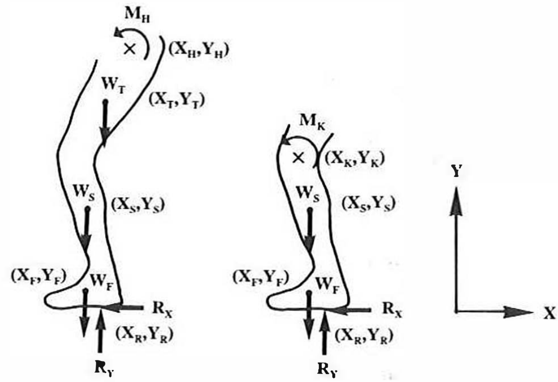
The participants wore shorts, and skin markers were placed at four sites on their left legs: 1) lateral side of fifth metatarsal head, 2) lateral malleolus, 3) center of lateral knee epicondyle and 4) greater trochanter of femur. They were filmed from the side at a focal distance of 7 m. A video-cassette recorder (Panasonic 8500) and a video camera (Hitachi FP-10) with a frame rate of 25 Hz were used. With a video position analyzer (FOR-A Company, VPA-1000), the coordinates for the markers in each frame were extracted from the video recordings.

The feet were placed symmetrically on a force plate (Kistler 9281 B, with four three-component force transducers) which measured the vertical, anteroposterior and lateral ground reaction forces during rising. The signals were channeled through amplifying units (Kistler 5006) to a microcomputer (Luxor, ABC 800), where they were A/D converted and stored. From the four original vertical force recordings, the application point in the horizontal plane for the summed vertical force was obtained. Combining these data with the video coordinates gave the appropriate sagittal moment arms with respect to the hip and knee joints. Dempster's anthropometrical data (7) were used to determine the segmental masses and their mass center locations.

The moments of force about the bilateral axes of motion of the knee (lateral epicondyle) and hip (superior part greater trochanter) were calculated from the initiation of movement until the body was in an upright standing position. The moments were calculated by multiplying each external force (body segment weight or horizontal or vertical reaction force) by its moment arm length (Fig. 3). All calculations were performed with the computer. The method for moment calculation was "semidynamic", i.e. it incorporated ground reaction forces as measured from the force plate but static contribution from body segments. Similar methods for moment calculation have been used earlier (4, 6, 8, 13, 16) and semidynamic methods have proved to give results very close to calculation with a fully dynamic method (10).

Electromyography and effort estimation

The muscular activity of the subjects' left vastus lateralis muscle was recorded (Devices M4, AC8) by means of a full-wave-rectified, low-pass-filtered and time-averaged electromyogram (linear envelope EMG). The low-pass time constant was 100 ms. Surface (Ag/AgCl) electrodes were placed on the skin over the muscle in the direction of the muscle fibers with an inter electrode distance of 2 cm. For control of artifacts, direct EMG was visualized in parallel on an oscilloscope (Tektronix RM565).



$$M_H = W_T(X_T - X_H) + W_S(X_S - X_H) + W_F(X_F - X_H) + R_X(Y_R - Y_H) + R_Y(X_R - X_H)$$

$$M_K = W_S(X_S - X_K) + W_F(X_F - X_K) + R_X(Y_R - Y_K) + Y_R(X_R - X_K)$$

Fig. 3. Calculations of the moments of force about the hip (M_H) and knee (M_K). R_X and R_Y are the horizontal and vertical components of the reaction force from the force plate. W_T, W_S and W_F are the segmental weights of thigh, shank and foot. (X_H, Y_H) and (X_K, Y_K) are the X and Y coordinates for the markers on the hip and knee joints. (X_T, Y_T), (X_S, Y_S) and (X_F, Y_F) are the X and Y coordinates for the center of gravity of the thigh, shank and foot. (X_R, Y_R) are the X and Y coordinates of the application point of the reaction force.

To quantify the muscular activity and to compare the activity between different subjects, EMG was normalized (12, 17). The EMG activity recorded during the test situation was related to EMG activity during an isometric maximum voluntary contraction (IMVC). A maximum isometric knee extension was performed in an exercise table with the knee flexed 60°. The EMG activity during the rising movements is expressed as a percentage of the EMG activity recorded during IMVC. Patients were excluded from EMG analysis since the maximal reference contraction was restricted by pain.

After each rise, the subjects assessed the subjective load using the Borg scale (3) (Table I).

Statistics

Since the paired differences were approximately normally distributed, the parametric *t*-test could be used.

RESULTS

Hip and knee moments

Fig. 4 shows the mean maximum moments, with a 95% confidence interval, for the hip and knee joints

Table I. *The Borg scale for estimation of effort*

6	
7	Very, very easy
8	
9	Very easy
10	
11	Fairly easy
12	
13	Somewhat hard
14	
15	Hard
16	
17	Very hard
18	
19	Very, very hard
20	

while rising with and without the flap seat. The subjects' mean peak hip moment was low both with (32.2 Nm) and without (35.8 Nm) the flap seat. The patients created a mean peak hip moment of 50.3 Nm when rising from the ordinary chair and this moment was reduced to 34.9 Nm when the flap seat was used. The decrease was significant at the $p < 0.005$ level. For both subjects and patients, use of the flap seat significantly reduced the mean peak knee moment for the subjects from 73.3 Nm to 41.3 Nm and for the patients from 52.7 to 33.0 Nm ($p < 0.001$).

Seat forces

The seat normal force N was divided into its vertical N_V and horizontal N_H components (Fig. 2*b*) to give some conception of how the force acted forwards and upwards. How the seat normal force N and its components varied with seat angle is shown in Fig. 5*a*.

The friction coefficient q determined the maximum friction force (at a given seat normal force) and the

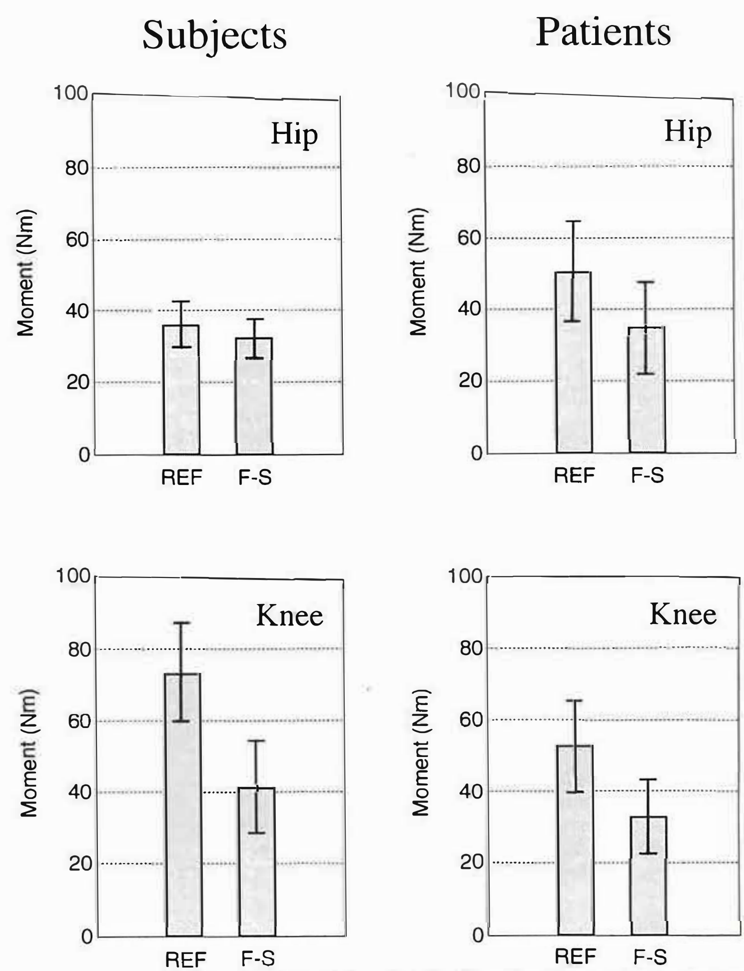


Fig. 4. Mean maximum hip and knee moments when rising with (*F-S*) and without (*REF*) the flap seat, 95% confidence intervals indicated.

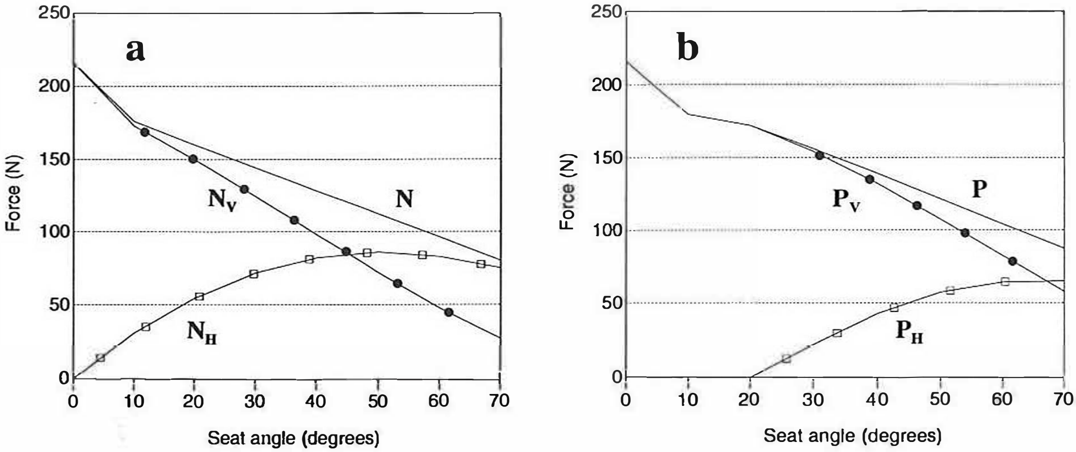


Fig. 5. (*a*) The seat normal force N and its vertical N_V and horizontal N_H components in relation to flap angle. (*b*) The effective seat force *P* and its vertical P_V and horizontal P_H components.

maximum angle between N and P. The friction coefficient was calculated to 0.41 (mean for 2 subjects) and the corresponding angle α to 22.3°. This angle and the following illustrations of the force P assume maximum utilization of the friction, but lower degrees of utilization are also, of course, highly possible and would give a smaller α and smaller P. The effective seat force P was also divided into its vertical P_V and horizontal P_H components (Fig. 2*c*) and these forces in relation to seat angle are shown in Fig. 5*b*.

Thigh muscle activity

The mean of the subjects' peak muscle activity in the vastus lateralis muscle when rising with and without the flap seat is shown in Fig. 6. The decrease of the mean caused by the flap seat was significant at a $p < 0.005$ level.

Estimation of effort

Each individual estimate of the effort of rising is shown in Table II. Most subjects estimated the rising equally easy with and without flap seat. All patients except one thought that rising without the flap seat was "somewhat hard" and use of the flap seat decreased the effort about four steps on the Borg scale.

DISCUSSION

It is clear that all patients felt that the flap seat facilitated the rising. This is of importance if the device is to be used at all. The decrease in estimated effort correlates well to the decrease in mean maximum hip and knee moment. The force from the seat helped the patients to decrease their knee moment by about 40%, which of course is of great importance for a patient

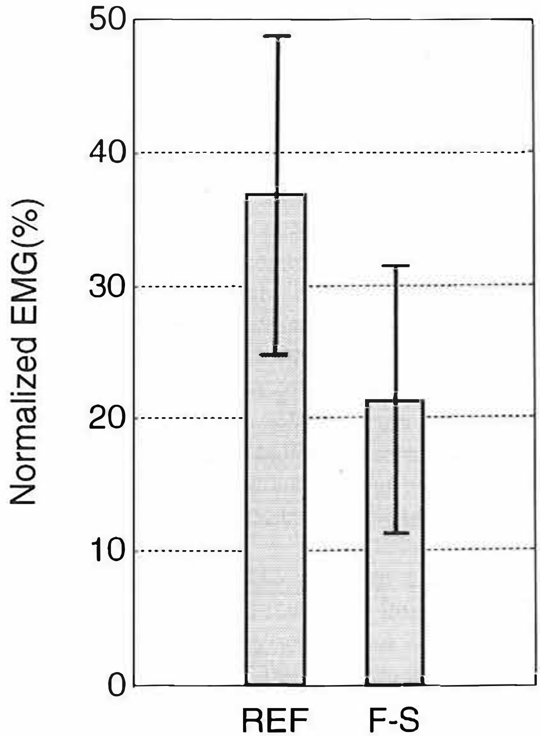


Fig. 6. Mean maximum muscle activity of the subjects' vastus lateralis with a 95% confidence interval, when rising with (*F-S*) and without (*REF*) flap seat. Amplitude normalized against activity during isometric maximum contraction.

Table II. *Individual estimations of effort using the Borg scale. REF represents rising without flap seat*

Subject	1	2	3	4	5	6	7	8	9	$\overline{X}$
Flap seat	6	11	12	8	7	7	8	8	8	8.3
REF	7	13	12	10	8	10	9	11	9	9.6
Patient	1	2	3	4	5	6	7	8		$\overline{X}$
Flap seat	9	9	10	8	9	10	9	10		9.2
REF	13	13	15	16	12	15	13	14		13.9

with knee osteoarthrosis. The hip moment was reduced by about 30%. The load reduction both makes it easier to rise from the chair and prevents further destruction of the affected joint. When joint loads for the patients and the subjects are compared it is clear that the patients distributed their load more equally between hip and knee. They did so by using a different technique, leaning the trunk more forward during rising. The flexing moment on the affected knee joint was thereby reduced and the moment on the hip increased. This was most obvious when rising without the flap seat.

Like many previous authors (e.g. 4, 10, 12), we used moment of force to indicate joint load. The linear and angular accelerations of the lower leg and thigh segments during rising from a sitting position were considered to be small and only little influenced by use of the flap seat. Inertial forces acting on the lower extremities would therefore cause only minor errors in calculation of hip and knee loading moments with a semidynamic model such as we have used. Semidynamic methods have also proved to give results very close to calculation with a fully dynamic method (10).

From the external view the patients appeared to rise symmetrically. On the other hand it is obvious that one cannot assume symmetry between the load on the affected and unaffected leg. Our calculations on hip and knee load for the patients should therefore be seen as an average load for the ipsilateral joints together, and as a method of studying total reduction in moment of force.

A flap seat like this, which not only pushes the patient upward, but also forward, may cause problems. In this study only patients with quite good balance and muscle strength participated, but patients with more pronounced muscle weakness and poor balance might easily stumble forward at the end of the movement. Fig. 5*a* shows that the vertical force component dominates at the beginning of the rise. At the end, however, (seat angles over 45°) the horizontal component is larger than the vertical. This observation, combined with the fact that the subject's center of gravity at that stage of rising is further forward, can make old and disabled people lose balance. Fig. 5*b* shows the force situation when the theoretically maximal friction force between subject and seat is added to the seat normal force. A comparison of Figs. 5*a* and *b* clearly shows that the effective seat force is larger for all seat angles when the friction force is utilized. The ratio between the vertical and the horizontal component is also more advantageous. This means that high friction between subject and seat is of great importance for optimal use of the device, and consequently slippery seats should be avoided.

The subjects used about 30% of the EMG reference level of the vastus lateralis muscle when rising without aid. This activity was significantly reduced when the flap seat was used. It is likely that the patients would have used a higher percentage of their maximum muscular activity when rising from the chair. A decrease in thigh muscle activity can be the factor that enables them to rise by themselves. Not only would the lower demand for muscular strength be beneficial, so would also a reduction of pain/discomfort due to a lower mechanical load on joint components.

This study shows that a flap seat like this can be a good help to people with knee osteoarthrosis. However, further development of the flap seat is desired to make it more suitable for weak and disabled people. Perhaps the most important thing is to make it possible to vary the seat force and thereby allow the user to adapt the force to bodyweight and fitness. With a too strong force the device can be dangerous and a weak force will not assist the rising as required.

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