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Lifting characteristics of functionally limited elders

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Abstract—Background. The purpose of this study was to investigate the lifting characteristics of elders with functional limitations using burden lifting smoothness, trunk angular momentum, and back and hip torque, and to correlate these characteristics with strength and functional measures.

Methods. Thirty elders (65–89 years old) consented to biomechanical analysis of lifting, gait, and chair rise, and to maximum isometric strength testing of the hip and knee extensors and shoulder flexors. Jerk, the rate of change of acceleration, is a measure of smoothness of motion. We calculated peak vertical jerk of the box at the beginning part of the lift. Momentum is the product of mass and velocity. HAT (head, arms, trunk) angular momentum was calculated during chair rise.

Results. Hip extensor strength correlated positively with box jerk, as did box jerk and peak trunk angular momentum between subjects. There was an inverse correlation between peak upper body angular momentum when lifting a box from floor to knee height, and gait speed normalized to body weight. There was a positive correlation between trunk angular momentum during lifting and trunk angular momentum during chair rise.

Conclusion. Stronger subjects used more peak vertical box jerk and more trunk peak angular momentum to lift a box from floor to knee height. Stronger subjects who used more HAT angular momentum during free speed chair rise also used more trunk peak angular momentum during the first phase of the lift, but lifting characteristics were independent of gait velocity. Weaker subjects used less peak momentum and peak jerk, choosing instead a more conservative, and apparently more stable, lifting strategy. Before counseling elderly patients on proper body mechanics for lifting, clinicians should assess strength and functional status. Weak elders should be taught a lifting strategy that allows them to maintain optimal balance, and to lift without jerking the load.

Key words: *angular momentum, balance, biomechanics, elderly, jerk, lifting, strength.*

INTRODUCTION

Lifting biomechanics have been studied extensively; though few published studies pertain to the elderly or

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attempt to relate lifting biomechanics with other functional measures. Although some authors have investigated the back muscles' extensor moment (1–7), much of the literature explores the back and the legs' interaction during lifting (8–14). These studies suggest that most subjects prefer to lift with a combination of back lift with straight knees, and leg lift with bent knees (15), but convert to back lift with increasing fatigue (16). No study to date reports jerk (the rate of change of acceleration) to quantify lifting strategy, and none have correlated lifting variables with other functional variables. Higher jerk implies less smooth, therefore less coordinated, movement. It is well known that the elderly are more prone to develop strength and functional impairments, which might predispose them to falls, often resulting in fracture (17). There is a high incidence of osteoporosis in the elderly population, where vertebral compression fracture could occur while lifting heavy loads or with jerky-type movements, such as opening a jammed drawer or window (18). Understanding the relationship of strength and balance, including chair-rise ability and strategy and gait velocity, should help clinicians determine a safe lifting strategy for elderly patients. Some clinicians now teach weak elders to use momentum producing, rapid lifting strategies, but the effect of such lifting strategies on safety and stability is unknown. Teaching an elderly person with limited knee motion or quadriceps weakness to use a leg lift strategy might threaten the person's stability, leading to a fall.

Patterson et al. (19) investigated the differences between experienced and inexperienced lifters. They concluded that experienced lifters exhibited more coordinated lifting strategies to decrease lower back stress, compared with inexperienced lifters. Flash and Hogan (20,21) used jerk to quantify movement smoothness during planar, horizontal arm movements. Because minimum jerk implies smooth, bell-shaped velocity profiles (20,21), minimizing mean-squared jerk during the movement gave an index of maximum smoothness. Minimum jerk while moving from one point to another, beginning at rest in one position and ending at rest in another position, has been studied extensively (20,21). Hogan and Flash state that higher cortical levels plan ideal movement trajectories and lower levels translate them into torque and momentum (21); muscle states define an equilibrium position for the limb (22). The muscle length–tension relationship, involving both reflexive mechanisms and mechanical properties, plays a significant role in postural control and movement (23).

It has long been recognized that lifting involves coordination as well as strength. Boston et al. (24) studied hip and knee angle coordination during repetitive squat lifting in subjects with chronic lower back pain and a control group. They reported that the movement patterns of the hip and knee were different between groups, with the control group finishing the movement of both joints at the same time, while the back pain group finished the joint movement segments at different times. Scholtz et al., (12,13) reported that beginning lifts from the same squat posture did not insure that a subject would lift with the same speed, and in the same manner, with increasing load. They found that subjects' peak posterior whole body center of mass (COM) velocity increased significantly with heavier loads. They concluded that this might aid in generating momentum to lift the load, but they only examined young (<53 years old), healthy subjects, who were able to tolerate postural stability-threatening momentum.

When rising from a chair, subjects with strength, balance, or functional impairments must use a strategy that controls momentum while maintaining balance and stability (25–30). Schenkman et al. (25) concluded that two possible strategies could be used to compensate for lower limb impairment: 1) position the upper body over the COM in a more favorable position over the feet or 2) use the upper body forward momentum to transfer into vertical momentum during the initial phase of arising, controlled by coordinated leg movement. Dynamic stability, or maintaining postural stability while the body is moving, is important for a momentum transfer strategy to be successful (25). Elderly subjects with moderate impairment used a momentum transfer strategy while maintaining stability (25). Weaker subjects have less capability to control for greater momentum and keep the base of support close to the COM for optimal stability (28). No study to date reports correlating sit-to-stand with lifting, therefore, we do not know if weaker subjects would use a momentum transfer strategy during lifting, as they do during chair rise.

Several groups investigated the relationship between age, strength, gait, and functional parameters. Many authors state that decreased gait velocity is a result of physical impairment (31–34). Winter et al. (35) stated that the elderly altered their gait pattern to be most stable and safe. It is unknown if, or how, the elderly alter their lifting strategy, and if the compensation could be predicted by assessing other functional measures.

The purpose of our study was to investigate the lifting characteristics of elders who are functionally limited

using vertical jerk of the load, trunk angular momentum, and torque at the hip and back, during unconstrained lifting. We determined the subjects' hip and knee extension strength impairment by hand-held dynamometry. We measured subjects' functional limitations during lifting, freestyle chair rise, and gait. Subjects were determined to be functionally limited by self-report using the SF36 physical function scale. Our general hypothesis was that weaker subjects would use less appropriate strategies than would stronger subjects during a controlled box-lifting task. Our specific hypotheses were:

1. Weaker subjects will compensate for muscle strength impairment by using more trunk peak angular momentum to lift a 5-kg box from floor-to-knee height than will stronger subjects.

2. Weaker subjects will use more vertical peak box jerk when lifting a 5-kg box from floor-to-knee height than will stronger subjects.

3. When lifting a 5-kg box from knee height to the point of maximum, whole body, vertical center of gravity (CG) displacement, trunk momentum will decrease as back and hip torque increase.

METHODS

Subjects

We studied a convenience sample of 30 subjects, 22 female and 8 male, who ranged in age from 65 years to 89 years (**Table 1**). The 30 subjects were selected from 96 elders who volunteered, signed a consent form, and completed the lift task. The subjects were first divided into three equal-sized groups: weak (1.0–10.0 kg), average (10.0–15.1 kg), and strong (15.2–28.0 kg) isometric knee extension strength as determined by hand-held dynamometer measurements. Buchner and deLateur (36) reported that knee extensor strength was representative of lower limb strength. We chose the first 10 subjects in each group who had complete data. We established three groups to include subjects with a wide range of strength values. The diagnosis of each subject was not a factor in group placement but diagnoses were similar between groups. Inclusion criteria included: at least 60 years of age, English speaking, able to use the telephone, able to walk indoors with or without assistive device (cane, walker) but not requiring human assistance, cognitively intact as determined by telephone questionnaire, and permission of subject's primary care physician. In a telephone screening interview, all subjects responded that they had one or more functional limitations on the SF36 physical function sub-scale—excluding the vigorous activity item (37). Exclusion criteria included: failure to pass an exercise tolerance test, uncontrolled hypertension or a blood pressure response exceeding 210/110 mmHg (assessed during baseline exercise tolerance test), history of a recent malignancy involving active chemotherapy or radiation therapy, significant coronary heart disease, chronic hepatitis, renal disease, diabetes, hyperthyroid or parathyroid dysfunction, congestive heart failure, significant cardiac arrhythmias, and rheumatoid arthritis, as provided by the subject's physician. Thirty subjects is the minimal acceptable number for a correlation analysis at p=0.05 for a power of 0.80, r=0.50. The power analysis was done before the subjects were selected. This study was approved by the institutional review board of Massachusetts General Hospital, Boston, MA.

Table 1.
Subject characteristics.

	Group 1		Group 2		Group 3	
	Mean	SD	Mean	SD	Mean	SD
Age	76.09	8.97	77.24	6.53	73.12	5.53
Height	1.59	0.07	1.60	0.10	1.66	0.10
Weight	65.50	10.37	68.65	9.18	69.70	11.39
Knee Strength	9.04	1.73	13.86	1.16	19.76	4.44
Hip Strength	10.59	4.05	11.97	3.49	14.43	6.49
Shoulder Strength	6.75	1.12	11.73	4.35	13.77	4.56

SD=standard deviation; height in m; weight and strength in kg.

Instrumentation

Instrumentation has been described in detail previously (25,38). Rigid arrays with light-emitting diodes

(LEDs) imbedded in them were attached to 11 body segments: head, trunk, pelvis, 2 arms, 2 thighs, 2 shanks, 2 feet, and 1 array was attached to the box. Kinematic data were collected with 4 Selspot (Selcom AB, Partille, Sweden) optoelectronic cameras with a viewing volume of 2×2×2 m, which tracks the LEDs. Each segment was modeled as a rigid body with 6 degrees of freedom—3 translations and 3 rotations (39). Telemetered Rapid Automatic Computerized Kinematic (TRACK) software (40) was used to calculate each segment's kinematic data. Accuracies in determining LED array positions are 1 mm for linear displacement and <1º for angular displacement. Two piezoelectric force plates (Kistler) measured ground reaction forces and moments.

Inertial parameters of each body segment (except the box) were derived from regression equations by McConville and Young (41,42). Kinematics of body segments and the box were measured by tracking the LED arrays. A pointing trial (39) was performed at the beginning of each test day to determine the position and orientation of the arrays relative to their respective body segments. Body segment momenta were computed from the body segment mass and COM velocity. Joint torques were calculated using inverse Newtonian dynamic analysis. Kinematic and force plate data were sampled at 150 Hz.

The box was a plastic crate with a mass of 0.5 kg. The dimensions were 0.33 m in width and length and 0.28 m in height, with side handles 21 cm above the bottom of the box. The box contained a 4.55-kg metal disc, which was placed on an aluminum pole in the center of the box. A quadrilateral Plexiglas array with four infrared LEDs was positioned at the lower left side of the box, 12 cm from the top (42). Box movement was measured as the displacement of its measured center of the array.

Procedure

Lift: Each subject stood with feet 30 cm apart at the heels, at the rear of the force plates. The box was placed with its corners in front of the subject's second toes. A table 94-cm high was placed 3 cm in front of the box. The subject was instructed to "lift this 10-lb box any way you like, to the table in front of you. Begin when I say 1-2-ready-go." After 1 practice trial, each subject completed 2 trials, which were sampled for 7 s.

Gait: Subjects walked independently without an assistive device, on a 10-m walkway, and completed at least 3 strides before entering the camera viewing area. The subject was instructed to "move forward in as straight a line as possible, walking at your own pace, as if you were taking a brisk walk in the park." Following several practice trials, each subject completed 2 trials, which were sampled for 3 s.

Sit-to-Stand (Chair Rise): The subjects were seated on an armless, backless chair at a height equal to their knee height, the distance from the medial tibio-femoral joint line to the floor. The feet were 10 cm apart at the medial foot borders, and the greater trochanters were 4 cm from the seat edge. The subjects' arms were folded with hands grasping their forearms, which were against the chest. Each subject was instructed to "rise up from the chair the way you usually do, after I say 1-2-ready-go. Then stand as still as possible until I say stop." Subjects were also instructed not to move their feet from the initial position or to use their arms. Each subject completed 1 practice and 2 trials, which were sampled for 7 s.

Hip and Knee Extension and Shoulder Flexion Strength: Strength was tested in the subject's home, by 1 of 2 examiners, within 2 weeks prior to functional testing. The subject was seated to test the knee, shoulder, and for hip abduction, and was standing, holding onto a chair backrest, to test hip extension. Maximum voluntary isometric contraction, measured with a hand-held dynamometer, was measured for a 3-s hold at 60º of knee extension, 90º of shoulder flexion, and 10º of hip extension. Two measurements were recorded after one practice trial. We added hip and knee extension strength values for data analysis because we were interested in the combined leg strength as well as individual muscle group strength. Intra-class correlation coefficient for hip extension was 0.9782, knee extension was 0.9754, and shoulder flexion was 0.9734. This demonstrates good test-retest reliability for the strength measurements.

The testers were trained in the testing protocol, which was documented in detail. The biomechanical measurement and digitizing were computerized ensuring reliability.

Data Analysis

Peak upper body angular momentum was recorded for two time frames: (*a*) time between initial box displacement from the floor-to-knee height, and (*b*) time between box displacement at knee height and maximum whole body vertical CG displacement.

Peak vertical box jerk was calculated during time between initial box displacement from the floor-to-knee height. All kinematic data were low-pass filtered at 6 Hz (43). Box displacement data were filtered using a mean

boxcar-smoothing window of 0.15 s. Successive time derivatives were used to calculate box velocity and acceleration. Accelerations were also mean smoothed with a boxcar window of 0.1 s, followed by another time derivative to calculate jerk. Smoothing parameters were chosen using fast Fourier transformation analysis to produce optimal jerk data, without compromising lower derivatives.

Peak back torque and hip torque were recorded during the time frame between box at knee height to maximum whole body vertical center of gravity displacement. Left and right three-dimensional hip torques were calculated separately, and then averaged for data analysis. We added hip and back torque for data analysis because both act in the same direction to lift the upper body. Hip torque extends the pelvis and back torque extends the trunk when lifting a box from the floor. We also did a within subjects time series correlation between back torque and upper body angular momentum during the same time frame. We analyzed the average of left and right hip torques. Head, arms, trunk, and pelvis (HAT) angular momentum at lift-off of freespeed chair rise is the combination of upper body angular momentum around a point midway between the hip joints. All momentum values were normalized to the subject's mass. Maximum vertical ground reaction force was used to determine time of lift-off because the chair was not on the force plate (44).

Freespeed gait velocity was determined by averaging the horizontal CG velocity over 3 s (first derivative of CG displacement). If there were more than one trial with complete data, we took the average values of the trials. Pearson product moment correlations were calculated to test the hypotheses, using the 6.1.3 version of the SPSS software (SPSS, Inc., Chicago IL).

RESULTS

There was no significant correlation between hip or knee isometric strength, and lifting momentum, r=0.1816, p=0.173 and r=0.0179, p=0.463, respectively. Although weaker subjects walked more slowly, and slower walking subjects used less trunk momentum, the first hypothesis was not supported. Raw data were reported in **Table 2.** During this first part of the lift, when the burden moved from floor to knee height, peak trunk angular momentum correlated directly with freespeed chair rise HAT angular momentum at lift-off, r=0.4998, p=0.006. There was an inverse correlation between trunk (HAT) angular momentum during the first part of the lift and gait speed normalized to body weight, r= −0.4803, p=0.004 (**Figure 1**). There was a positive partial correlation between gait speed and hip + knee strength, r=0.4199, p=0.023, and gait speed and knee strength, r=0.3701, p=0.041, when both strength and gait speed were body weight adjusted. **Figure 2** shows a curvilinear gait speed/strength relationship: gait speed increased as strength increased until the 30 kg hip + knee strength threshold, beyond which strength increase did not result in faster gait speed.

Table 2.
Raw data.

Variable	Mean	SD
Chair trunk HAT ang. mom.	0.07 kg*m/s	0.07
Trunk mom. lift floor to knee	0.34 kg*m/s	0.17
Trunk mom. lift knee to CG	0.77kg*m/s	0.42
Hip torque lift knee to CG	60.19 N*M	12.45
Back torque lift knee to CG	71.18 N*M	23.82
Av. gait speed	110.27 cm/s	20.50
Jerk lift floor to knee	1,830.62 cm/s^3	627.48

SD=standard deviation, ang.=angular; mom=momentum; Av.=average; CG=center of gravity.

Contrary to hypothesis 2, weaker subjects did not use more jerk when lifting the box from floor to knee height. In fact, there was a positive correlation between hip strength and vertical peak box jerk, r=0.3467, p=0.003 (**Figure 3**). There was also a positive correlation between peak jerk and trunk peak angular momentum during the first part of the lift, r=0.3868, p=0.017 (**Figure 4**).

Box displacement and whole body vertical CG displacement are displayed in **Figure 5.** During a typical elder's lifting trial, there was a steep rise in vertical box displacement, but the peak displacement was always greater than the final resting position on the table. A similar pattern existed for vertical CG displacement. This excessive vertical displacement "overshoot," and the fact that some subjects swung the box to one side while lifting (**Figure 5**), suggests excessive work is performed to prevent collision with the table top.

A smooth, minimum jerk lift should generate bell-shaped velocity profiles (20–22). We found three types of velocity profiles (**Figure 6**). The most common, in 53 percent (n=30) of the trials, had two peaks. The first velocity peak corresponded approximately to the time that the box reached knee height. The second velocity

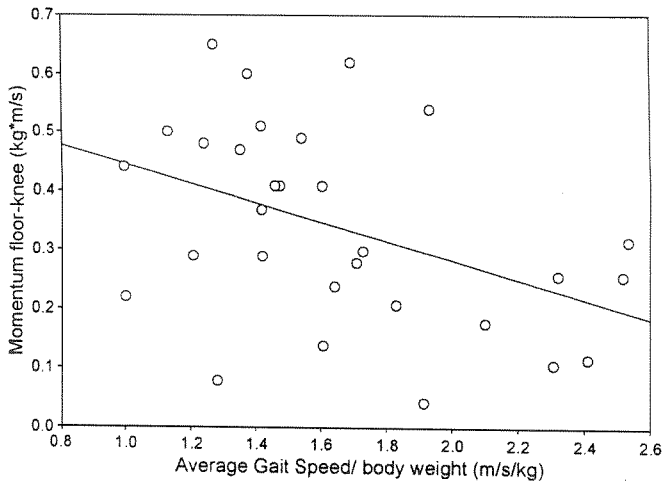

Figure 1.
Average gait speed/body weight versus upper body angular momentum during box lifting from floor to knee height.

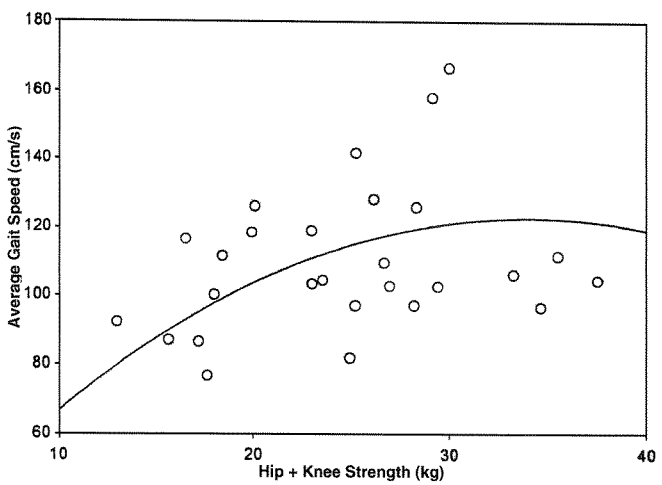

Figure 2.
Hip + knee isometric strength versus average gait speed, derived from a quadratic polynomial regression.

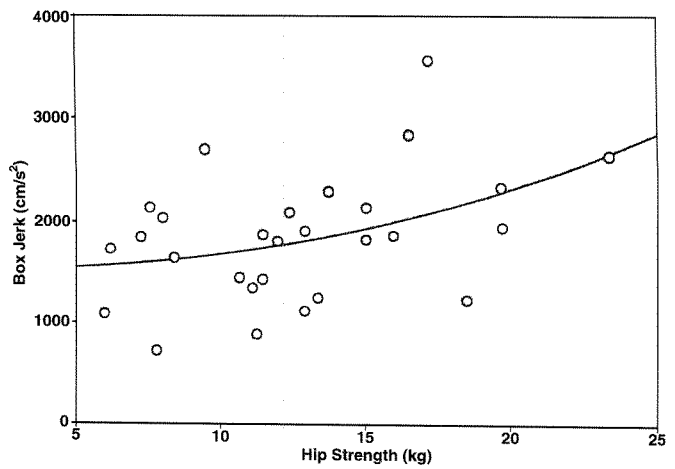

Figure 3.
Hip strength versus vertical box jerk during box lifting floor to knee, derived from a quadratic polynomial regression. Jerk values begin to rise as strength increases above 5 kg, and continue to rise nonlinearly with increasing strength.

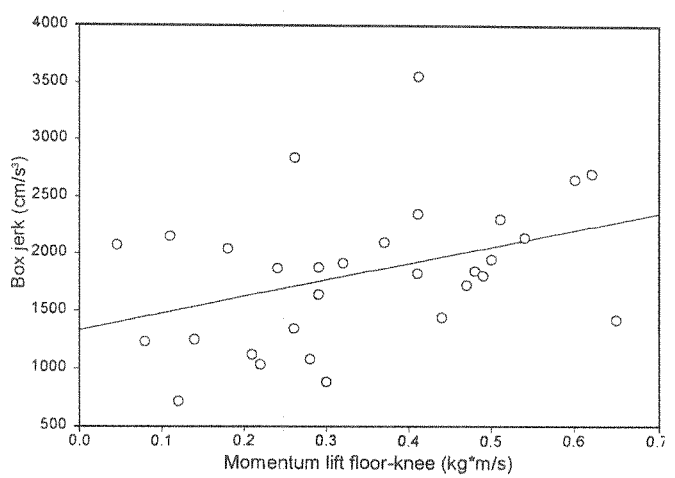

Figure 4.
Trunk angular momentum versus vertical box jerk during box lifting from floor to knee height.

peak occurred approximately 500 msec before the time of maximum vertical box displacement. In 36 percent of the trials (20 trials), there was a single, short duration peak, which occurred between the times that the box reached knee height and the time of maximum vertical CG displacement. In 11 percent (6 trials) there was a single, long duration peak (approximately 1.5 s) with the beginning and end times corresponding to those of the peaks in the bimodal profiles.

In support of hypothesis #3, a within-subjects time series correlation between back torque and upper body angular momentum revealed a negative correlation in 22 of 30 subjects, with an r to Z transformation average of r= −0.178, standard error =0.215, for all 30 subjects. There was, however, a positive across-subjects correlation between peak hip torque and peak momentum, r=0.3601, p=0.025 (**Figure 7**), and between peak hip + back torque and peak momentum, r=0.3423, p=0.032. There was a significant positive correlation across subjects, between maximum isometric strength and peak hip and back torque (**Table 3**).

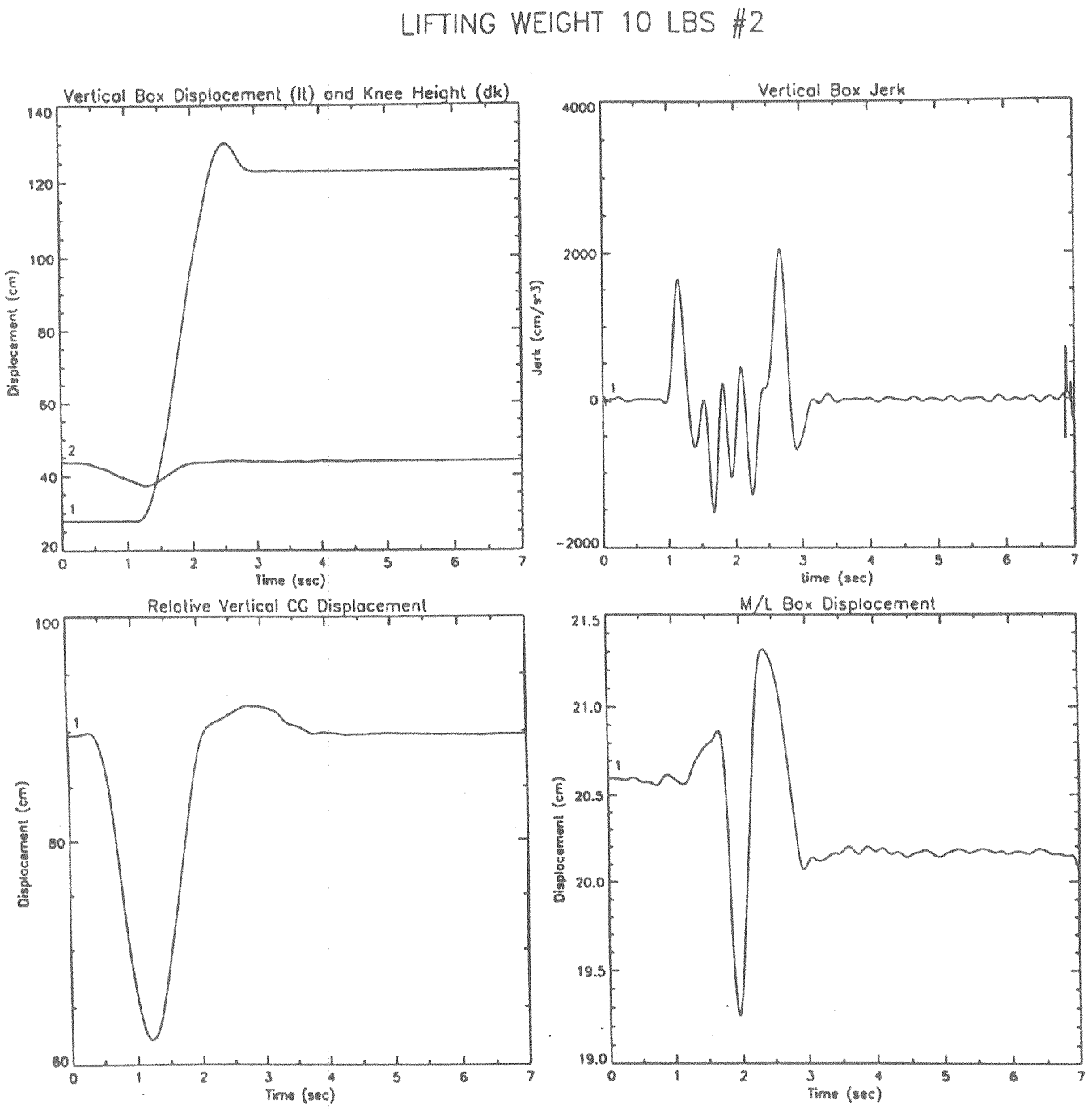

Figure 5.
Raw data for vertical and mediolateral box displacement, and whole-body vertical center of gravity displacement. Vertical box displacement does not begin at zero because this line was offset to represent the top of the box rather than the measured array position near the bottom, so that the intersection of the two lines is when the top of the box reaches knee height. Vertical box jerk (top right graph) was examined from the initial displacement time to the time the top of the box reached knee height (top left graph).

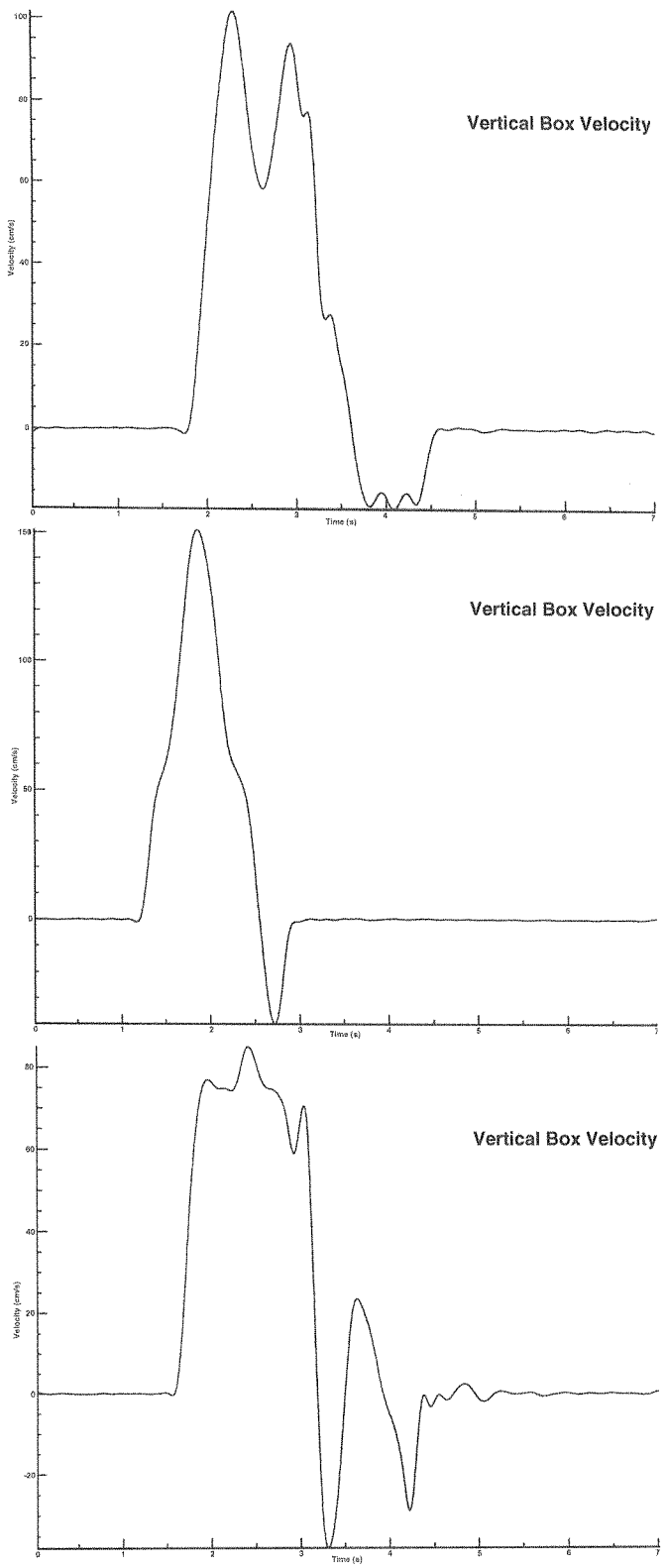

Figure 6.
Three profiles of vertical box velocity.

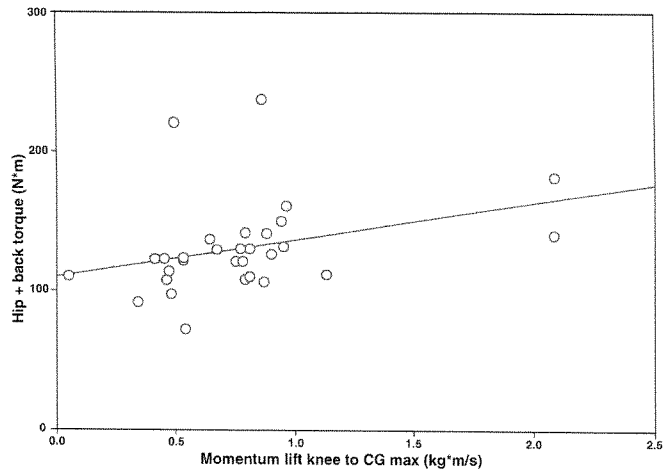

Figure 7.
Trunk angular momentum versus hip + back torque during box lifting from knee height to maximum vertical center of gravity displacement.

Table 3.
Strength measures for three different lifting strategies, by box velocity profiles.

	Velocity Profiles		
Strength Values	**Bimodal**	**Single Sharp Peak**	**Single Wide Peak**
Knee Extension			
Mean	8.97	15.90	12.61
SD	3.64	5.94	4.61
Range	7.00-22.50	8.50-28.50	7.60-20.50
Shoulder Flexion			
Mean	8.90	13.13	8.53
SD	3.05	4.93	2.47
Range	4.95-18.65	5.75-20.35	6.05-13.05

Extensions and flexions in kg; SD=standard deviation.

DISCUSSION

Different from the reported characteristics during chair rise (28,30), weaker subjects did not compensate for their strength impairments by increasing momentum during the initial phase of lifting. Stronger functionally limited elders used more momentum and more jerk when lifting a box from floor to knee height than weaker subjects. Stronger subjects were apparently more willing to threaten their postural stability by using more dynamic lifting mechanics than were weaker subjects. Similarly, our subjects who used more trunk angular momentum to lift also used more HAT angular momentum to rise from a chair. Scarborough et al. (45) also reported that weaker

subjects used less HAT linear momentum during chair rise lift-off, suggesting that weaker subjects used a less destabilizing strategy to rise from a chair. During lifting, Toussaint et al. (10) concluded that there is normally a close relationship between the moment caused by the vertical ground reaction force and the net back and leg joint moments, such that the ground reaction force vector direction helps stabilize the subject. Weaker subjects in our study apparently used a more stable lifting strategy to maintain balance, thus decreasing their trunk torques and momenta.

No studies to date report functional lifting characteristics of elderly subjects. In single-joint motor control studies (20–23), less jerk means smoother movement, which in turn is more energy efficient. Most prior studies used arm movements and reported bell-shaped velocity profiles during graceful movement. Most of our elderly subjects (53 percent) used bimodal, but bell-shaped, vertical velocity profiles while lifting the box (**Figure 6**). The two vertical velocity curves probably represent two movement segments, with the velocity decrease occurring as the box decelerates prior to being conservatively lifted higher than the final platform height. During the final movement segment, the box must be translated forward to be placed on the table, which could independently decrease the vertical box velocity, causing two distinct movement segments. Subjects using a bimodal velocity profile had less shoulder flexion and knee extension strength (**Table 3**) than did the subjects with the two other velocity profiles. The first peak appears to correspond to leg and trunk impulses. The second peak appears to correspond to shoulder and elbow flexion to lift the box further to reach table height. The second lifting style, using the single long duration peak (11 percent of subjects), could represent a merging of, or smoother transition between, the vertical and horizontal movement segments. The subjects using the single, long duration peak velocity strategy had greater mean knee extensor strength than the subjects with bimodal velocity profile, but no difference in shoulder flexion strength. The 36 percent of subjects using a single sharp peak curve (bell-shaped profile) represent the smoothest, strongest lifters. These subjects had the highest shoulder flexion and knee extension strength (**Table 2**). Thus, weaker subjects tended to use less jerk by dividing the lift into two distinct movement segments, while stronger subjects merged the entire lifting task into one smooth movement.

We used peak jerk, instead of mean-squared jerk of the entire lift, to avoid the time-dependent differences the latter would engender and because we were interested in the absolute amount of jerk during the first phase of the lift. In our study, greater peak lifting jerk correlated with more trunk momentum, suggesting that stronger subjects had more dynamic postural stability.

All subjects overshot, then lowered the box to the table (**Figure 5,** top left). Relative whole body vertical CG displacement also showed this overshoot pattern (**Figure 5,** bottom left). This is not unlike the pattern of foot and whole body CG displacement during stair ascent first reported by Zachazewski et al. (46). These findings suggest that this overshoot in vertical displacement is a result of conservative motor planning: undershooting the table (or stair) would result in a collision and potentially a fall, while overshooting the final position merely wastes energy.

Because the entire lift took <1 s, predictive feed forward control must play at least a part in movement coordination. Predictive controls rely on memory, are highly context specific, and are expressed in integrated actions serving a final goal (47). In testing subjects grasping different weights between thumb and index finger and lifting them to a specified point, Johansson and Westling (48) found that with a constant weight, adequate force development was predictively programmed before lift-off from the table. In lifting with unexpected weight changes between lifts, the force rate profiles were programmed in relation to the previous weight. With lifts programmed for heavier weight, the high load and grip force rate showed an excessive overshoot and high grip force peak. Schneider and Zernicke (49) tested four subjects moving their non-dominant arm from a low to a high target for 100 trials. They found that movement times decreased, hand paths became more symmetrical, and jerk cost (mean-squared jerk) was significantly lower for practiced movements. No change occurred in the last 75 trials. Our subjects had only one practice lift and two lifting trials during which data were collected. This lack of learning opportunities could also explain the lift overshoot.

Torque and momentum time-histories within subjects were inversely related during most subjects lift (22–30 subjects) from the time that the box reached knee height to the time of upright standing, as we hypothesized (**Figure 8**). For a constant energy increase of a conservative system with no power loss, torque and angular velocity are inversely related (50). The human body is not a conservative system, in that energy is added to and removed from the system depending in part on lifting strategy. There was, by contrast, a positive correlation

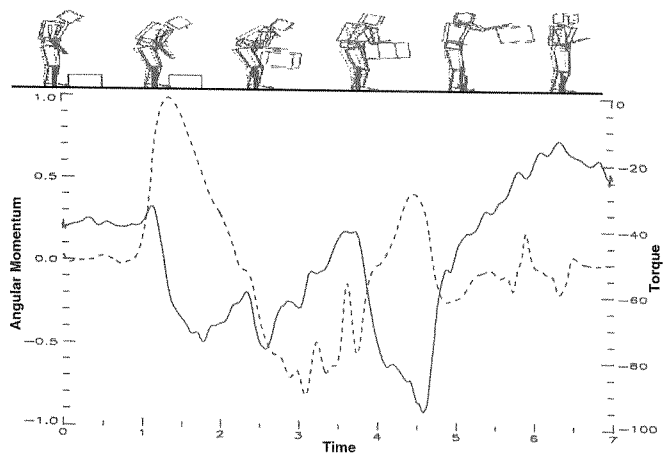

Figure 8.
Time-series data for upper body angular momentum (dashed line) and back torque (solid line) in a typical elder. Flexion momentum and torque are shown as positive values on the graph; note the inverse momentum-torque relationship during the lift. The androids above the time-history graph reveal whole body kinematics at 1.36, 2.21, 3.08, 3.48, 4.25, and 4.55 s during this lift.

between hip and back peak torque and upper body peak angular momentum among subjects during the lift (**Figure 7**). It appears that the initial lifting torque is used to generate momentum, which must be dissipated and controlled during the latter part of the lift. Stronger subjects generated more back and hip peak torque and more trunk peak angular momentum than did weaker subjects during lifting (**Table 4**), suggesting that weaker subjects chose a more stable lifting strategy.

Buckner and deLateur (36) described a theoretical lower limb strength threshold, below which gait speed curvilinearly decreased and above which no further gait speed improvement was realized. To our knowledge, these are the first data (**Figure 2**) to empirically support Buckner and deLateur's model (36). Our data suggest that lower limb strength is the limiting factor in very weak elders' gait, but once adequate strength (about 30 kg hip + knee maximum isometric voluntary contraction) obtains, other factors contribute to gait speed variations.

Table 4.
Peak trunk angular momentum during lifting and isometric hip and knee isometric strength correlations to peak hip and back torque during lifting.

Torque	Hip		H+B		Back	
	r	p	r	p	r	p
Peak momentum	0.3601	0.025	0.3423	0.025	0.2857	0.063
Hip Strength	0.5904	0.000	0.6068	0.000	0.5132	0.002
H+K Strength	0.7662	0.000	0.7353	0.000	0.5890	0.000

H+B=hip plus back; H+K=hip plus knee.

Gait velocity does not appear to be a predictor of a person's willingness to destabilize their CG during lifting. **Figure 1** shows that subjects who walk faster tend to use less momentum in lifting, but only if walking speed is normalized to body mass. **Figure 2** shows that weaker subjects walked more slowly than stronger subjects, but the relationship was nonlinear and occurs without body mass normalization. It has been reported previously (45) that weaker subjects use less momentum for chair rise. Our data also reveal that chair-rise trunk momentum is more related to lifting momentum, and therefore is a better predictor of lifting strategy, than is gait.

We tested our subjects using a freestyle, submaximal lift. This lack of controlling the lifting style might account for the low correlation between muscle strength and trunk momentum during the first part of the lift. Controlling the type of lift (i.e., stoop or squat), requiring maximal effort, or changing the foot position might change the strength-momentum correlations, but at the expense of standardization and generalizability. Most subjects chose a combination of stoop- and squat-lifting, thus supporting our choice of not forcing a particular lifting style to be followed.

Suggestions for further study include analyzing the effects of a strength and conditioning program on lifting measures in the elderly (17,36,51). Further investigation of the effects of repetitive lifting practice on the box displacement and velocity profiles may provide more insight into movement smoothness and how particular lifting strategies are chosen.

There are several clinical implications of this study. Before counseling elderly persons on proper body mechanics for lifting, clinicians should assess strength and functional status. Assessing a person's strength and chair rise strategy could provide information on the individual's functional stability, which could influence instruction on his/her lifting strategy. That subjects select a lifting strategy that does not substitute trunk momentum to compensate for lower limb weakness, suggests that counseling people to substitute rapid lifting to ease the burden transfer may imperil the subject's balance and potentially threaten his or her postural stability. Weak elders should be taught a lifting strategy that allows them to maintain optimal balance, such as keeping a wide base of support, lifting without jerking the load.

CONCLUSIONS

• Stronger subjects used more peak vertical box jerk and more trunk peak angular momentum to lift a box from floor-to-knee height than did weaker subjects. Weaker subjects used less peak momentum and peak jerk, choosing instead a more conservative, more stable lifting strategy.

• Stronger subjects who used more HAT peak angular momentum during free speed chair rise, also used more trunk peak angular momentum to lift a box from floor-to-knee height.

• Lifting characteristics are independent of gait velocity.

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REFERENCES

1. Dolan P, Mannion AF, Adams MA. Passive tissues help the back muscles to generate extensor moments during lifting. J Biomech 1994;27:1077–85.
2. Schultz AB, Haderspeck-Grib K, Sincora G, Warwick D. Quantitative studies of the flexion-relaxation phenomenon in back muscles. J Orthop Res 1985;3:189–97.
3. Bogduk N. A reappraisal of the anatomy of the human erector spinae. J Anat 1980;131:525–40.
4. McGill SM, Norman RW. Effects of an anatomically detailed erector spinae model on L4/L5 disc compression and shear. J Biomech 1987;20:591–600.
5. Toussaint HM, deWinter AF, deHaas Y, et al. Flexion relaxation during lifting: implications for torque production by muscle activity and tissue strain at the lumbo-sacral joint. J Biomech 1995;28:199–210.
6. Gracovetsky S, Farfan H. The optimum spine. Spine 1986;11:453–71.
7. Dolan P, Adams MA. The relationship between EMG activity and extensor moment generation in erector spinae muscles during bending and lifting activities. J Biomech 1993;26:513–22.
8. deLooze MP, Toussaint HM, vanDeen JH, Demper HCJ. Joint moments and muscle activity in the lower extremities and lower back in lifting and lowering tasks. J Biomech 1993;26:1067–76.
9. Schipplein OD, Traftmow JH, Andersson GBJ, Andriachi TP. Relationship between moments at the L5/S1 level, hip and knee joint when lifting. J Biomech 1990;23:907–12.
10. Toussaint HM, vanBaar CE, vanLangen PP, et al. Coordination of leg muscles in back lift and leg lift. J Biomech 1992;1279–89.
11. deLooze MP, Kingma I, Bussman JBJ, Toussaint HM. Validation of a dynamic linked segment model to calculate joint moments in lifting. Clin Biomech 1992;7:161–9.
12. Scholtz Jp, Millford JP, McMillan AG. Neuromuscular coordination of squat lifting: effect of load magnitude. Phys Ther 1994;75:119–32.
13. Scholtz JP, McMillan AG. Neuromuscular coordination of squat lifting II: individual differences. Phys Ther 1994;75:133–44.
14. Vrahas MS, Brand RA, Brown TD, Andrews JD. Contribution of passive tissues to the intersegmental moments at the hip. J Biomech 1990;23:357–62.
15. Bejjani FJ, Gross CM, Pugh JW. Model for static lifting: relationship of loads on the lumbar spine and knee. J Biomech 1984;17:281–6.
16. Hagen KB, Sorhagen O, Harms-Ringdahl KB. Influence of weight and frequency on thigh and lower trunk motion during repetitive lifting employing squat and stoop techniques. Eur J Appl Physiol 1993;67:291–7.
17. Topp R, Mikesky A, Wigglesworth J, et al. The effect of a 12 week dynamic resistance strength training program on gait velocity and balance of older adults. Gerontologist 1993;33:501–6.
18. Myers ER, Wilson SE. Biomechanics of osteoporosis and vertebral fracture. Spine 1997:22:25S–31S.
19. Patterson P, Congleton J, Koppa R, Hutchinson RD. The effects of load knowledge on stresses at the lower back during lifting. Ergonomics 1987;30:539–49.
20. Hogan N, Flash T. Moving gracefully: quantitative theories of motor coordination. TINS 1987;10:170–4.
21. Flash T, Hogan N. The coordination of arm movements: an experimentally confirmed mathematical model. J Neurosci 1985;5:1688–703.
22. Hogan N. An organizing principle for a class of voluntary movements. J Neurosci 1984;4:2745–54.

23. Hogan N. Planning and execution of multijoint movements. Can J Physiol Pharmacol 1988;66:508–17.
24. Boston JR, Rudy TE, Mercer SR, Kubinski JA. A measure of body movement coordination during repetitive dynamic lifting. IEEE Trans Rehab Eng 1993;1:137–44.
25. Riley PO, Schenkman ML, Mann RW, Hodge WA. Mechanics of a constrained chair-rise. J Biomech 1991;24:77–85.
26. Schenkman M, Berger RA, Riley PO, et al. Whole body movements during rising to standing from sitting. Phys Ther 1990;70:638–48.
27. Yoshida K, Hiromutsu I, Fumo I. Motion analysis in the movements of standing up from and sitting down on a chair. Scand J Rehab Med 1983;15:133–40.
28. Hughes MA, Schenkman ML. Chair rise strategy in the functionally impaired elderly. J Rehabil Res Dev 1996;33(4):409–12.
29. Schultz AB, Alexander NB, Ashton-Miller JA. Biomechanical analysis of rising from a chair. J Biomech 1992;25:1383–91.
30. Alexander NB, Schultz AB, Warwick DN. Rising from a chair: effects of age and functional ability on performance biomechanics. J Gerontol 1991;46:m91–8.
31. Gabell A, Nayak USL. The effect of age in variability in gait. J Gerontol 1984;662–6.
32. Imms FJ, Edholm OG. Studies of gait mobility in the elderly. Age Aging 1981;10:147–56.
33. Elbe RJ, Hughes S, Higgins C. The syndrome of senile gait. J Neurol 1992;239:71–5.
34. Gibbs J, Hughes S, Dunlop D, et al. Joint impairment and ambulation in the elderly. J Am Geriatric Soc 1993;41:1205–11.
35. Winter DA, Patla AE, Frank JS, Walt SE. Biomechanical walking pattern changes in the fit and healthy elderly. Phys Ther 1990;70:340–7.
36. Buchner DM, deLateur BJ. The importance of skeletal muscle strength to physical function in older adults. Ann Behav Med 1991;13:91–8.
37. Ware JE, Sherbourne CD. The MOS 36-item short form health survey (SF36). I. Conceptual framework and item selection. Med Care 1992;30:473–83.
38. Krebs DE, Wong D, Jevsevar D, et al. Trunk kinematics during locomotion activities. Phys Ther 1992;72:505–14.
39. Riley PO, Mann RW,Hodge WA. Modeling of the biomechanics of posture and balance. J Biomech 1990;23:503–6.
40. Antonsson EK, Mann RW. Automatic 6-DOF kinematic trajectory acquisition and analysis. ASME J Dynam Syst Meas Contr 1989;111:31–9.
41. Grierson YM. Creation of a three dimensional human body model for dynamic analysis (Masters thesis). Massachusetts Institute of Technology; 1992.
42. Leupongsak N, Krebs DE, Olsson E, et al. Hip stress during lifting with bent and straight knees. Scand J Rehabil Med 1997;29:57–64.
43. Angeloni C, Riley PO, Krebs DE. Frequency content of whole body gait kinematic data. IEEE Trans Rehab Eng 1994;2:39–46.
44. Riley PO, Popat RA, Krebs DE. Whole body momentum analysis of failed chair rise. IEEE Trans Rehab Eng. In press.
45. Scarborough DM, Quadriceps muscle strength and dynamic stability in elderly persons (Masters thesis). MGH Institute of Health Professions; 1996.
46. Zachazewski JE, Riley PO, Krebs DE. Biomechanical analysis of body mass transfer during stair ascent and descent of healthy subjects. J Rehabil Res Dev 1993;30(4):412–22.
47. Johansson RS, Cole KJ. Grasp stability during manipulative actions. Can J Physiol Pharmacol 1994;511–23.
48. Johansson RS, Westling G. Coordinated muscle commands adequately and erroneously programmed for the weight during lifting task with precision grip. Exp Brain Res 1988;71:59–71.
49. Schneider K, Zernicke RF. Jerk-cost modulations during the practice of rapid arm movements. Biol Cybern 1989;60:221–30.
50. Beer FP, Johnson ER, editors. Mechanics for engineers. New York, NY: McGraw-Hill Co; 1988. p. 841,933–6.
51. Judge JO, Underwood M, Genosa T. Exercise to improve gait velocity in older persons. Arch Phys Med Rehabil 1993;74:400–6.

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