

Comparison of seat, waist, and arm sit-to-stand assistance modalities in elderly population

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Abstract—The ability to perform a sit-to-stand (STS) motion is important for ambulatory adults to function independently and maintain daily activities. Roughly 6% of community-dwelling older adults experience significant difficulties with STS, a major risk factor for institutionalization. While mechanical STS assistance can help address this problem, full dependence on STS assistance provided by devices such as lift chairs can lead to atrophy of the leg muscles. We investigated the mechanics of assisted STS motion in order to better understand how load-sharing STS mechanisms may facilitate STS motions while still requiring activation of the leg muscles. Experiments were conducted with 17 nondisabled older adults performing unassisted and assisted STS rises with grab bar, arm, seat, and waist assistance. Each mode of rise was evaluated based on a subject questionnaire and key biomechanical metrics relating to stability, knee effort reduction, and rise trajectory. Results show that the seat- and waist-assist modes provide statistically significant improvements in stability metrics and reductions in required knee torques over unassisted rises and bar assistance. The assists most preferred by the subjects were the seat and bar assists. Overall, our results favor a seat-assisted STS modality for non-clinical applications and indicate further testing of this modality with a clinical population.

Key words: assistive device, assistive robotics, assistive test bed, biomechanics, chair rise, elderly, load sharing, mobility impairment, seat assist, sit-to-stand.

INTRODUCTION

Rising from a chair is a basic requirement of maintaining independence for ambulatory older adults [1].

Difficulty with the sit-to-stand (STS) motion is common among elderly people, affecting more than 6 percent of community-dwelling older adults [2] and over 60 percent of nursing home residents [3]. STS is considered the most mechanically demanding functional task undertaken during daily activities [4]. Furthermore, there is a significant unmet need for assistance with STS [5], potentially leading to institutionalization and reduced functionality in performing activities of daily living [6].

Presently, there are a number of assistive STS devices either available commercially or in development. Commercial devices include passive supports, such as grab bars and standing frames that provide stability as users rise, and active supports, such as lift cushions, lift chairs, and powered standing devices [7]. In the research and development phase, STS devices include walker systems with powered standing aids [8–10], a powered handrail [11], and a moveable bed and support bar system [12].

Given the range of commercial STS devices, it is unclear which types of devices are the most appropriate

Abbreviations: CoM = center of mass, CoP = center of pressure, MT = momentum transfer, NSERC = Natural Sciences and Engineering Research Council of Canada, RMS = root-mean-square, STS = sit-to-stand.

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for persons who require partial assistance. While powered lift devices are frequently used, concern exists among rehabilitation professionals that habitual use of lifts may contribute to accelerated muscular degeneration due to muscle disuse. It is, therefore, important to better understand the biomechanics of the STS process in order to design improved devices that can achieve appropriate trade-offs between the functional goal of enabling users to stand safely and the long-term therapeutic goal of maintaining and building muscular strength.

In this article, we present the design of an STS test bed intended to evaluate how four common modes of assisted STS, representing the types of assistance provided by various commercial and in-development STS assistive devices, affect the biomechanics of older adults rising from a chair, as assessed using a variety of metrics previously used in related studies [13–16].

In this primary study, we evaluate STS assistance on a population of physiologically normal older adults who have sufficient muscular endurance to participate in the comparative study; based on the results, we can plan more focused experiments with a more clinically relevant population. We compared the biomechanics of both unassisted and assisted STS rises using four different assist modes: bar, arm, waist, and seat (illustrated in the “Sit-to-Stand Test Bed” section). Based on metrics from previous unassisted STS studies [13–16], we evaluated each assistance mode to determine which assists (1) provide the greatest amount of static stability to the subject, (2) provide the greatest amount of dynamic stability to the subject, (3) result in the greatest reduction in knee extensor effort required to rise while still sharing with the subject part of the knee load required to rise, (4) enable a subject to follow a momentum transfer (MT) rise strategy (often clinically preferred and targeted in rehabilitation programs), and (5) are most preferred by the subjects. We anticipate that the waist and seat assistance modes will induce the largest deviations from the unassisted rise on all metrics, since they provide the largest levels of assistance. We also anticipate that the bar- and arm-assist modes will have relatively little effect on the stability and load-sharing measures, even though overall stability may well be enhanced through the contact between the hand and the environment (this enhanced stability is not directly assessed by the force plate-based measures).

METHODS

Experiment Design

We evaluated the questions outlined previously using the following key biomechanical metrics, all drawn from previous studies reported in the STS biomechanics literature. With respect to the stability measures, two biomechanical measures are primarily used in the literature to measure the stability of a STS rise: the displacement of the whole body center of mass (CoM) to measure static stability (postural balance) [15,17] and the displacement of the foot center of pressure (CoP) from the foot center at seat-off time to measure dynamic stability (postural stability) [14].

Static Stability

Static stability was assessed using the separation between subject CoM and his or her ankle at the time when the load-sharing assistance ended or at seat-off time in the cases where load-sharing assistance was not provided. A smaller separation indicates greater static stability [15,17].

Dynamic Stability

The measure for dynamic stability was determined by the location of the foot CoP with respect to the center of the foot at the time when the load-sharing assistance ended or at seat-off time in the cases where load-sharing assistance was not provided. Dynamic (postural) stability is maximized when the foot CoP is centered between heel and toes [14].

Extensor Effort

Extensor effort was assessed by the peak knee torque as calculated from sensor data using a standard four-segment serial linkage model (foot, leg, thigh, head-arm-trunk), with lower knee torques indicating less knee extensor effort required to rise [16]. We assessed both the absolute level of knee torque as well as the peak knee torque ratio, calculated according to **Equation (1)**:

$$\text{Peak Knee Torque Ratio} = T_{k\text{-assisted}}/T_{k\text{-unassisted}}, \quad (1)$$

where $T_{k\text{-assisted}}$ = assisted peak knee torque and $T_{k\text{-unassisted}}$ = unassisted peak knee torque.

Momentum Transfer Strategy

To assess closeness to the MT strategy as described by Scarborough et al. [13], the peak trunk flexion for each of the assisted STS rises was compared with the peak trunk flexion of the unassisted STS rises using the MT strategy.

Subjective Preference

To determine subjective preference, subjects completed a postexperiment questionnaire (**Table 1**) for all of the modes of STS assistance.

Sit-to-Stand Test Bed

A test bed (**Figure 1**) was built for this experiment that included a seat platform; a commercially available grab bar [7]; and seat-, waist-, and arm-assist modes based on existing STS assistance modes. All assistance mechanism motions were user-controlled through a “dead man” switch that must be depressed to enable motion. The primary function of the non-load-bearing arm-assist mode is to provide stability and trajectory guidance. The controllers for the waist- and seat-assist mechanisms are designed to reduce peak knee torque and encourage subjects to rise using their available strength. The control scheme for the waist- and seat-assist modes is a velocity-based controller, designed such that if users apply a knee torque greater than 35 percent of the unassisted torque, the assistance moves at 100 percent of the nominal velocity for an unassisted rise. This threshold is based on the finding of Hughes et al. [16] that young adults use approximately 35 percent of their maximum

available knee torque for STS; for patients for whom an STS movement requires virtually all of their strength, 35 percent of this effort should feel significant but not overwhelming. If a subject applies between 35 and 0 percent of the unassisted torque, the velocity is ramped down proportionally, with 100 percent velocity if 35 percent knee torque is applied and 0 percent velocity if 0 percent knee torque is applied. The 100 percent nominal velocity for an unassisted rise was selected such that, in pilot tests, the waist- and seat-assist modes could bring users to a standing position with a rise time between 2 and 4 s, which corresponds with natural STS speeds in older persons [18]. This velocity-based control scheme therefore provides a level of assistance that depends on each subject’s own weight and rising pattern.

Subjects

Seventeen community-dwelling nondisabled elderly subjects (11 male and 6 female) over the age of 60 were recruited. Subjects were selected if they were able to rise unassisted from a chair and did not have any of the following contraindications: known musculoskeletal or neuromuscular conditions that would limit their ability to rise from a chair, balance disorders, osteoporosis, recent significant injury or treatment, recent hip or knee replacement, current rehabilitation care, or current fainting or dizzy spells [19].

Protocol

Age, weight, height, and anthropometric data were collected and are summarized in **Table 2**. Data for one subject were recorded but not used because of a data collection error during her trials. Anthropometric data included lengths of the foot, shank, thigh, upper body, upper arm, and forearm found by palpation at joints to estimate rotation points. The test bed seat height was pre-adjusted to 80 percent of the knee height, measured from the floor to the left medial tibial plateau [13]. In this study, 80 percent height was used as a standard seat height [20]. Furthermore, we wanted thighs approximately parallel to the ground in the seated position (to achieve a full 90° of thigh motion from seated to standing position) and this was better achieved at a seat height of 80 percent. Small angular displacement sensors (described in the “Test Bed Data Collection” section) were attached to subjects along the midline of the left anterior shank and thigh and the midline of the anterior chest wall at the approximate CoM of each body segment

Table 1.

Postexperiment questionnaire. For each assist, subjects chose response from 4-point Likert scale: (1) disagree, (2) somewhat disagree, (3) somewhat agree, and (4) agree.

Question Number	Question Text
1	I felt stable when using this assist.
2	I was able to rise with this assist using the <i>same motion</i> as used during the unassisted rise.
3	I was confident that I would not fall while rising using this assist.
4	I was able to rise smoothly with this assist.
5	I felt comfortable in terms of forces placed on my body while rising using this assist.
6	I was able to rise with this assist using <i>less effort</i> than the effort required to rise unassisted.

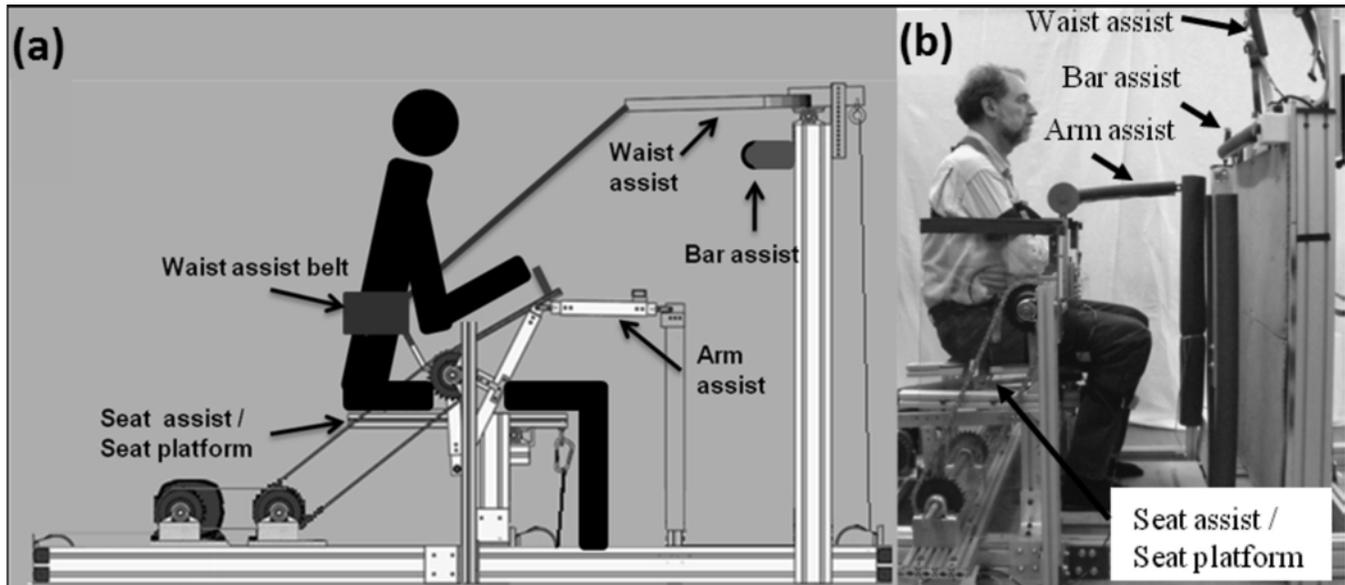


Figure 1.

Sit-to-stand test bed. (a) Person ready for waist-assisted rise. Other three assistance modes are also labeled. (b) Person ready for unassisted rise.

Table 2.

Summary of demographic and anthropometric data from 16 nondisabled older adult subjects.

Data	Mean \pm SD
Age (yr)	71.0 \pm 5.8
Mass (kg)	70.1 \pm 7.7
Height (cm)	172.0 \pm 6.4
Length (cm)	
Foot	24.5 \pm 2.0
Shank	38.1 \pm 2.8
Thigh	47.1 \pm 2.8
Trunk	44.6 \pm 4.2
Upper Arm	25.9 \pm 2.6
Forearm	25.4 \pm 1.3
Seat Height (cm)	39.6 \pm 2.0

SD = standard deviation.

[21]. Motion sensors were mounted in the anterior plane to avoid sagittal plane shear and over the segment's CoM to increase mounting stability and minimize movement due to skin distortion from joint flex. The feet were sock-covered and placed in parallel on a single force plate (described in "Test Bed Data Collection" section), shoulder-width apart. Each shank was positioned at approximately 18° flexion with respect to the vertical plane, approximating normal foot placement at the start of a constrained chair-rise motion [4].

To ensure that subjects uniformly engaged with the assistance modes during assisted STS, subjects were given instructions for each rise mode. Subjects were asked to sit and maintain their trunk and head in an upright position. They were then asked to look straight ahead and, upon a "go ahead" cue, rise to an erect position in one of the following five modes:

1. Without assistance at a self-selected speed, using a typical MT STS strategy [22] as demonstrated by the experimenter.
2. Holding the grab rail for the duration of the rise to increase stability but requiring the subject to provide the knee torque necessary to stand up (similar to Bahrami et al. [15]).
3. Using the arm-assist mechanism to guide the rise trajectory during the STS motion but requiring the subject to provide the knee torque to stand up.
4. With the waist-assist belt-type mechanism providing partial assistance.
5. With the pivoting seat assistance providing partial assistance.

Only power-assisted modes 4 and 5 reduced the required knee torque from the subject. An STS rise was considered successful if the feet remained fixed during the rise and if subjects maintained contact with the assistance

mechanism during rise. A set of five rises was recorded for each mode of STS. Upon completion of each mode, subjects answered the questionnaire for that mode.

In each STS mode, except for the arm- and bar-assist modes, subjects rose with arms crossed to ensure that all rising forces were generated by the legs or the assistance devices. The unassisted STS was performed, followed by the four assisted modes. A modified randomized block design was used to counterbalance the four assisted modes of STS, with the arm-assist modes being either the first or last mode to facilitate switching between assistance modes. For the assisted trials, subjects were permitted to perform several practice trials to familiarize themselves with the assistance mechanism. Subjects were informed that the waist- and seat-assist control scheme would slow the assistance mechanism if subjects did not use at least 35 percent of the knee torque required for unassisted standing.

Test Bed Data Collection

The lift load supported by the seat was measured with an S-type load cell (model PT 4000–300lb, Pressure Transducers Ltd; Auckland, New Zealand) located underneath the seat. A 6-axis force plate (model OR6–7-1000–5571, Advanced Mechanical Technology Inc; Watertown, Massachusetts) located beneath the subject's feet measured the ground reaction forces and located the fore and aft CoPs. Two S-type load cells (model PT4000–50lb, Pressure Transducers Ltd) measured the load applied by the hip belt to the waist of the subject. Kinematic data were collected using motion sensors (model MTx-49A53G25, Xsens Technology Inc; Enschede, the Netherlands) (static root-mean-square [RMS] error: $<1^\circ$, dynamic RMS error: $<2^\circ$), which provided three-dimensional linear acceleration, angular velocity, and roll-pitch-yaw orientation data. Sensor drift was negligible over the short duration of each trial. Bergmann et al. [23] showed that mean motion sensor orientation measurements were strongly correlated with joint angle measurements from an optical tracking device (correlation range: 0.93–0.99, maximum joint ranges of motion: 49° – 92°), and we verified the angle measurement accuracy using an inclinometer; in addition, we verified the accuracy of the motion sensor during the rise against a video recording of a single subject.

For each trial, 10 s of force and kinematic data were collected, starting 1 s before the “go ahead” cue (all completed STS motions took less than 10 s). The data were

collected at 50 Hz and digitally filtered with a zero-delay, bidirectional, fourth-order, low-pass Butterworth filter at a cut-off frequency of 5 Hz [24]. Post hoc, the accelerometer data were gravity compensated based on the measured orientation of each motion sensor, and the force data were time-shifted to synchronize with the accelerometer data.

Data Analysis

Biomechanical Model and Data Treatment

The four-link rigid-body model used was based on Mak et al. and Kuo [24–25]. The approximate body-segment masses, CoM locations, and moments of inertia were calculated using generalized anthropometric coefficients [21]. Bilateral symmetry about the sagittal plane was assumed [15]. The joint forces and torques were recursively calculated using the Newton-Euler inverse dynamics method [26]. For the bar- and arm-assist modes, the four-link biomechanical model was extended to a six-link model by separating the head/arms/trunk segment into three segments comprising the head and trunk, upper arm, and forearm. To obtain the kinematics of the upper arm and forearm during the experiment, an additional motion sensor was attached to the upper arm on the anterior plane, along the midline, and over the approximate CoM of the upper arm. For the arm-assist mode, we assumed 90° forearm with respect to the upper arm because of the constraint of the arm rest. For the bar-assist mode, we calculated the forearm angle and CoM position based on the fixed position of the hand (on the bar) and the location of the elbow (derived using the upper-arm motion sensor).

The maximum trunk flexion was determined as the difference between the initial trunk angle reading and the peak trunk angle reading. The trunk flexion and thigh angle readings were used to determine the motion time. The start and end of motion were defined as the time at which the trunk angular velocity exceeded 0.1 rad/s [19] and the thigh extension angular velocity dropped below 0.1 rad/s, respectively.

To enable comparisons across subjects, data were normalized as follows: knee torque by the product of body height and mass, foot CoP and body horizontal CoM by total foot length, and the motion time linearly scaled between 0 and 1 [15]. The biomechanical metrics were averaged over each set of five trials, except in two cases in which only four trials were available because of

procedure or data errors for one trial. For the postexperiment questionnaire, the scores reported by subjects for all six questions were grouped for each assistance mode and combined for the 16 subjects.

Statistical Analysis

We performed a repeated-measures analysis of variance on the biomechanical metrics, followed by post hoc Bonferroni correction. A one-sample *t*-test was conducted to determine whether the peak knee torque ratio (**Equation (1)**) for each mode of assisted STS was significantly greater than 35 percent. A Friedman nonparametric repeated-measures analysis was applied to the postexperiment questionnaire data, followed by post hoc Wilcoxon signed-rank tests to identify differences between assistance modes. There were a total of 48 statistical tests; thus, the significance level was set at $\alpha = 0.05/48 = 0.001$ throughout.

RESULTS

The data were analyzed to address the five metrics described in the “Experiment Design” section. **Figure 2** presents a summary of the results.

Static Stability

The CoM and ankle separation in the seat- and waist-assist modes were both significantly smaller than in the unassisted STS ($p < 0.001$). The waist-assist mode CoM and ankle separation were also significantly smaller than for all other assisted STS rises ($p < 0.001$). The seat-assist mode CoM and ankle separation was significantly smaller than for the bar- and arm-assist modes ($p < 0.001$).

Figure 3 shows the horizontal projection of the CoM trajectory for a representative subject performing each of the five modes of STS. For all of the assists, the total-body

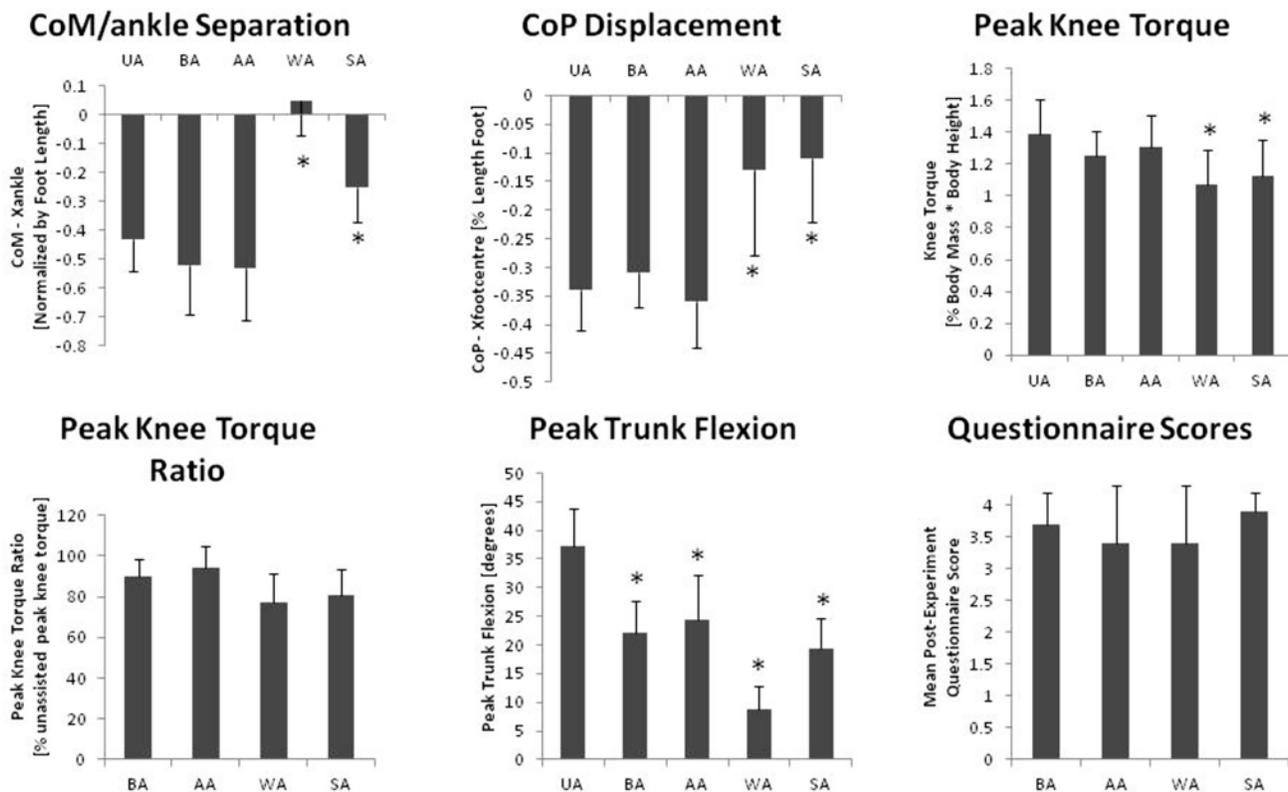


Figure 2.

Results from key biomechanical metrics and postexperiment questionnaire from all subjects for five modes of sit-to-stand (STS). Error bars indicate standard deviation. *STS modes with results that are significantly different from unassisted (UA) mode results. AA = arm-assist, BA = bar-assist, CoM = center of mass, CoP = center of pressure, SA = seat-assist, WA = waist-assist.

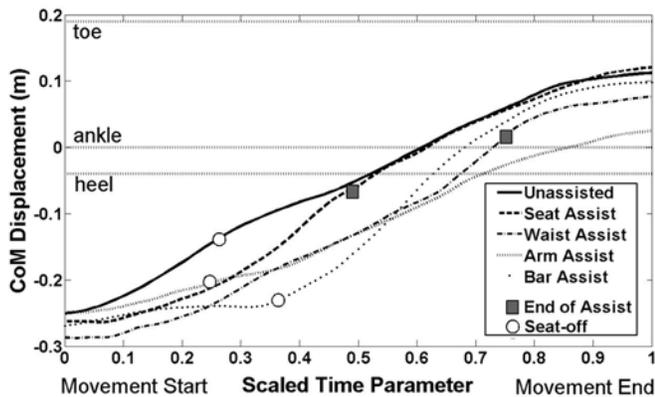


Figure 3.

Trajectory of horizontal projection of total-body center of mass (CoM) of representative subject for five modes of sit-to-stand (STS). For unassisted, arm-assisted, and bar-assisted STS rises, CoM/ankle separation is measured at seat-off time (circle). For seat-assisted and waist-assisted STS rises, CoM/ankle separation is measured at assistance end time (square).

CoM is monotonically increasing (i.e., moving forward). For the load-sharing waist- and seat-assist modes, the assistance ended later in the trajectory compared with the seat-off time for the other trajectories. Thus, the CoM at the time of full loading was closer to the ankle for the load-sharing assists.

Dynamic Stability

The CoP displacement with respect to the center of the foot in the seat- and waist-assist modes were both significantly smaller than all other STS modes ($p < 0.001$). Thus, the waist- and seat-assist modes were found to be the most dynamically stable modes of assisted STS, indicating that subjects were less likely to have a “step” or “sit back” failure [27] when using these assistance modes than with the other assistance modes. **Figure 4** shows the trajectory of the horizontal projection of the CoP for a representative subject.

Knee Extensor Effort and Load Sharing

Both the waist- and seat-assist modes reduced peak knee torques relative to the unassisted STS ($p < 0.001$). Despite the fact that both waist- and seat-assist modes produced similar mean reductions in peak knee torques, only the waist-assist mode resulted in a statistically significant reduction compared with the arm-assist mode

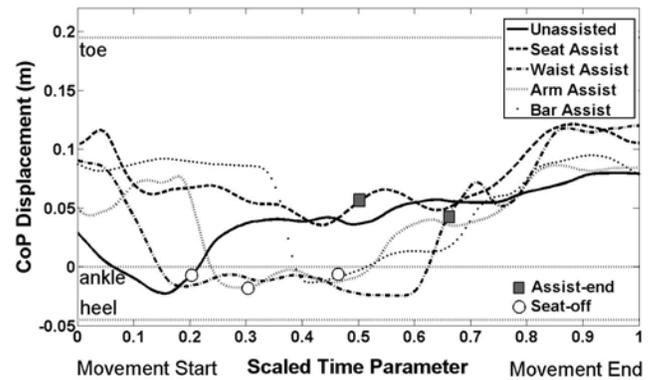


Figure 4.

Trajectory of foot center of pressure (CoP) of representative subject for five modes of sit-to-stand (STS). For unassisted, arm-assisted, and bar-assisted STS rises, CoP displacement from foot center is measured at seat-off time (circle). For seat-assisted and waist-assisted STS rises, CoP displacement from foot center is measured at assistance end time (square).

($p < 0.001$). **Figure 5** shows the trajectory of the knee torque for a representative subject.

As required by the extensor effort (load-sharing) criterion (see the “Sit-to-Stand Test Bed” section), the average peak knee torques generated in all four of the assisted STS modes was greater than 35 percent of average peak knee torque of the unassisted rise, with the waist-assist mode facilitating the lowest torque ratio at 77 percent of the peak unassisted knee torque (the seat-assist mode was slightly higher at 81%). Thus, both the waist- and seat-assist modes are able to significantly reduce the knee torque required to stand compared with unassisted STS.

Momentum Transfer Strategy Proximity

Significant differences in peak trunk flexion angle were detected between the unassisted STS and all of the assisted STS modes ($p < 0.001$). The waist-assist mode peak trunk flexion was significantly lower than seen in the other three assistance modes ($p < 0.001$). **Figure 6** shows the trajectory of the trunk angle for a representative subject performing each of the five modes of STS.

No assistance mode was able to closely replicate the MT strategy. The low peak trunk flexion suggests that all of the assisted STS rises promote a dominant vertical rise strategy [13], with the waist-assist mode providing the least amount of MT.

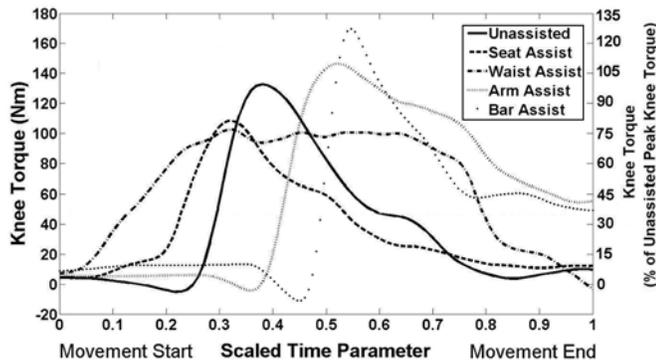


Figure 5. Knee torque trajectories of representative subject for five modes of sit-to-stand.

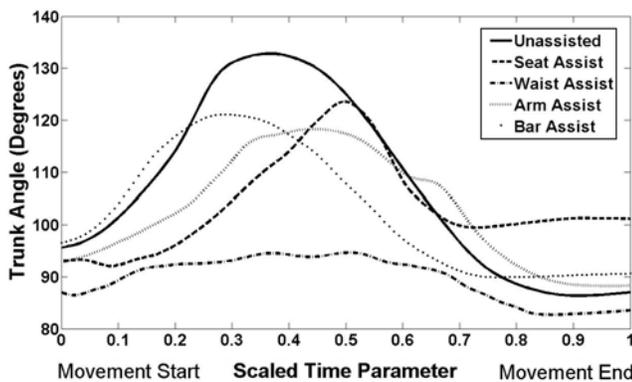


Figure 6. Trunk angle trajectories of representative subject for five modes of sit-to-stand.

Postexperiment Questionnaire

On the postexperiment questionnaire, the seat- and bar-assist modes had scores significantly higher than the scores for the arm- and waist-assist modes ($p < 0.001$). No difference was detected between the bar- and seat-assist modes (Figure 2).

DISCUSSION AND CONCLUSIONS

Comparisons between the assisted and unassisted STS modes enabled us to identify characteristic differences in the biomechanical metrics we chose to assess the motions. Overall, although the details of the trajectories

were considerably different between the various assistance modes, neither the bar- nor the arm-assist modes produced significant changes in most of the assessed parameters, with the exception of peak trunk flexion. In contrast, the seat- and waist-assist modes did result in significant changes with respect to the measures of static and dynamic stability and knee extensor effort, generally leading to changes in the direction that would be considered desirable when designing an STS assistance device, although both resulted in significant deviations from the MT approach that therapists often ask their patients to attempt to emulate (significantly more-so with the waist-assist mode). The postexperiment questionnaire showed that the most-preferred STS modes were the seat- and bar-assist modes. Overall, it appeared that the seat-assist mode may represent the best compromise—it induces a lower deviation from the MT strategy than the waist-assist mode (and comparable with the other assisted modes) while generating desirable changes in the other three metrics, as well as the highest subjective acceptability score. We therefore recommend that the seat-assist mode be evaluated further as a basis for future improvements in STS device designs.

This study's conclusions are potentially limited by our use of nondisabled older adults as subjects rather than older adults with functional limitations, who would better represent the ultimate target users of an STS device. However, in order to collect comprehensive data across repeated trials with a range of assistance devices, we required a subject pool who had sufficient strength to perform the required number of rises. As we proceed to refine the design of a future STS device, we will seek to target a more appropriate clinical population.

We also acknowledge that some of the biomechanical metrics we used are arguably more appropriate for unassisted standing than assisted modes in which the subjects make contact with the environment at points other than their feet. While development of more specialized metrics is likely warranted, we selected metrics that have been used previously in standing studies and would allow us to draw comparisons with previously published studies. For example, the lack of improvement in static stability in the bar-assisted rise is consistent with Bahrami et al. [15], who showed a strong deviation from the MT strategy for rising with a bar support. The knee torque reduction by the seat-assist mode is consistent with findings by Wretenberg et al. [28], who used a spring-loaded flap seat to reduce peak knee torque from 73 to 41 Nm

($p < 0.001$). Furthermore, the smaller peak trunk flexion found in the seat-assist mode is consistent with Bashford et al. [29], who found that rising with a lift chair resulted in a peak trunk flexion 6.1° less than the peak trunk flexion when rising without a lift chair.

We believe that this comparative study of multiple STS assistance modes provides useful information on the biomechanical differences induced by each of these modes that can be used to guide future STS device design efforts.

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Study design: J. Jeyasurya, A. Hodgson, H. F. M. Van der Loos, E. A. Croft.

Acquisition of data: J. Jeyasurya.

Analysis of data: J. Jeyasurya, E. A. Croft.

Statistical analysis: J. Jeyasurya, A. Hodgson.

Biomechanical analysis: J. Jeyasurya, H. F. M. Van der Loos.

Drafting of manuscript: J. Jeyasurya.

Critical revision of manuscript for important intellectual content:

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Study supervision: E. A. Croft, H. F. M. Van der Loos, A. Hodgson.

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Participant Follow-Up: The authors do not plan to inform participants of the publication of this study.

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