Abstract

Sit-to-stand transfer mechanics in healthy older adults: a comprehensive investigation of a portable lifting-seat device

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Purpose: To evaluate lower extremity mechanics and muscle activation associated with the sit-to-stand transfer using a portable lifting-seat device and to compare these data to an unassisted transfer in healthy young and older adults. *Methods*: Bilateral lower extremity and low back musculature electromyography, three-dimensional leg and trunk motion, and ground reaction forces were recorded from 10 young (mean age = 25) and 10 older (mean age = 69) adults during five trials of (i) no assist and (ii) assisted transfers. Data were time normalized to represent the period of seat-off to standing. Peak sagittal plane joint angles, moments, and muscle activity profiles were calculated. Analysis of variance models was used to test for main effects and interactions ($\alpha = 0.05$). *Results*: Trunk, hip, and knee angles were significantly reduced and dorsiflexion increased with assisted transfer ($p < 0.05$). Peak hip and ankle joint moments were reduced ($p < 0.05$) and no change found in knee moments ($p > 0.05$). Peak muscle activity was lower during the assisted transfer ($p < 0.05$). Seat device effects were similar between age groups. Older adults used higher relative muscle activation. *Conclusion*: Variables indicative of sit-to-stand functional demand were reduced with lifting-seat device use. Data provide a framework for future recommendations on product prescription, use, and research pertaining to the advancement of adaptive seating.

Implications for Rehabilitation

Hip and trunk mechanical demands, and muscle activation were reduced with portable lifting seat device use.

Greater ankle dorsiflexion was found with portable lifting seat device use, suggesting this range of motion should be considered when prescribing this device.

Healthy older and younger adults used similar knee and trunk joint mechanics yet older adults completed the sit-to-stand trials with greater lower extremity and low back muscle activation.

Keywords: Electromyography, lift-seat device, lower extremity, mechanics, older adults, sit-to-stand, rehabilitation

Introduction

For decades, rising from a seated position has been recognized as an activity of daily living that allows individuals independence in achieving many household chores, activities, and personal hygiene tasks [1–4]. The sit-to-stand transfer is one of the most demanding functional tasks that individuals undertake during daily life. At the hip joint, contact pressures between the femur and the pelvis are higher than during walking, jogging, or jumping [5]. At the knee joint, Hughes et al. [6] found that older individuals with functional impairments (i.e. inability to descend four stairs reciprocally or stand up from sitting at a 0.33 m seat height) utilized close to 78% of available knee extensor strength to complete the sit-to-stand transfer from a chair set to knee joint height whereas younger individuals required approximately 34%. In addition, Mizner et al. [7] found knee extensor moments (internal moments) to be approximately twice the amplitude compared with walking in those individuals with knee osteoarthritis who received a total knee replacement. No other activities of daily living place this mechanical demand on the human body. Given these demands, it is not surprising many individuals have difficulty with this task. Sit-to-stand has been studied in many contexts and may include the manipulation of numerous constraints (i.e. speed of rising [8,9], seat height [10,11], and foot position [12,13] using many outcomes (i.e. kinematics, kinetics, electromyography (EMG), and time characteristics). Sit-to-stand mechanics differ very little between older and younger adults when no significant impairments exist [6,14,15]; however, various populations of individuals have been studied to understand the limitations on sit-to-stand ability imposed by aging and/or disease [6,16–19]

Despite numerous studies that examine external and internal constraints on sit-to-stand movement characteristics in individuals with limitations in this activity, investigations of adaptive seating designs to overcome these constraints have been lacking. Specifically there is minimal information on how these devices change the mechanical demands (i.e. joint loading and muscle activation) associated with standing up.

In many instances, individuals with lower limb impairments are unable to perform a safe and effective sit-to-stand transfer from a fixed-seating system. Studies have shown that raised seat heights reduce the mechanical demands (i.e. lower sagittal plane hip and knee moments and reduced quadriceps activation amplitudes) when compared with a normal seat height [10,19– 22], but having a fixed-raised seat is not always practical for many environments of daily living. Systems have been designed to provide assistance to elevate the height of the chair from a standard height to a raised height. Early work on these designs focused on booster [23] or spring-loaded flap seats [24], although it was argued that these designs challenge balance, forcing the patient to stand up using abnormal movement mechanics [23,25]. Positive experiences, however, have been reported and largely comprise the evidence of lifting-seat effectiveness. Bashford et al. [25] found that 75% of the individuals rated easier transfers as a result of using an ejector mechanism, a result similar to that presented by another study [23]. Health care professionals, scientists, and industry have joined together to develop advancements in adaptive seating systems that aim to assist the sit-to-stand transfer [26]. These designs include systems without mechanical assists (e.g. adding arm rests, increasing seat height), systems with mechanical assists (e.g. Booster or ejector chairs), and those systems that can lift, tilt, recline, or rock.

To our knowledge, while investigations exist to support the use of non-mechanical assist techniques, extrapolated from the determinants of sit-to-stand [27], few studies have investigated the mechanical demands (i.e. biomechanics and muscle activation) of using seating devices to mechanically assist the sitto-stand transfer. Munro et al. [20,28] analyzed both the mechanics (angular kinematics and kinetics) and muscle activation patterns of individuals with rheumatoid arthritis as they transferred from sit-to-stand using an ejector chair. In general, participants rated their perceived exertion much lower when the ejector mechanism was employed. Trunk and knee angular displacement was significantly reduced but knee joint moments at seat off were not with the ejector mechanism compared with standard height [20]. Earlier quadriceps and tibialis anterior muscle activation onsets and longer activity durations (quadriceps) were found with ejector mechanism use as well. In addition, either no change or increased muscle amplitudes were found when rising with ejector assistance [28]. In contrast, Wretenburg et al. [23] found that the use of the spring-loaded flap seat reduced both external hip and knee flexion moments and vastus lateralis muscle activity. While comprehensive, how these findings translate to other lifting-seat designs, such as those that do not provide a horizontal thrust (eject or boost), remain unknown.

Therefore, the purpose of this investigation was to evaluate lower extremity and trunk motion, moments of force, and muscle activation associated the sit-to-stand transfer using a portable mechanical device to raise the seat height that does not provide a vertical or horizontal thrust and compare these data to an unassisted transfer in healthy young and older adults. It was hypothesized that joint angles, moments of force, and muscle activity, as a measure of mechanical demand, will be reduced with the assisted sit-to-stand and these differences will be similar between healthy younger and older adults.

Materials and methods

Participants

Ten young adults (five female and five male) and 10 older adults (five female and five male) were recruited from the general community. All participants were included if they self-reported that they were healthy, had no fracture or previous lower extremity injury other than a sprain or strain, no cardiovascular/ respiratory disease, neurological disorders (i.e. stroke, Parkinson's disease, myocardial infarct, and arrhythmias), or known lower extremity musculoskeletal disorders (i.e. osteoarthritis) that would affect their ability to rise from a chair. Written informed consent was obtained in accordance with the Institutional Research Ethics Board.

Procedures

All individuals completed the Lower Extremity Functional Scale (LEFS) [29] and participants' height and mass were recorded. Participants were familiarized to the sit-to-stand protocol with no device and with a power lifting seat device using accompanying device instructions, (Power Seat UPE-P100Ex 24VDC, Uplift Technologies Inc., Dartmouth, NS, Canada). Seat height was adjusted to knee height. Participants practiced the sit-to-stand movement45 times from the armless, backless seat design until comfort with the movement was reported. The experimental setup is illustrated in Figure 1.

Participants were prepared for skin surface electrode application consistent with current guidelines as suggested by the International Society of Electrophysiology and Kinesiology and SENIAM (Surface electromyography for the Non-Invasive Assessment of Muscles). Skin preparation included light shaving and abrading with 70% alcohol wipes. Surface electrodes (3M, Red Dot, Repositionable Monitoring Electrodes, St. Paul, MN) were placed in a bipolar configuration over tibialis anterior (TA), vastus lateralis (VL), vastus medialis (VM), biceps femoris (LH), gluteus maximus (GM) bilaterally, and over the left erector spinea (ES), 3 cm lateral to third lumbar spinous process as previously shown [30,31]. Muscle palpation and a series of isometric contractions for specific muscle groups were used for EMG validation and gain adjustment. Signals were amplified using two AMT-8 (Bortec, Inc., Calgary, Alberta, Canada), eight-channel EMG systems (Input Impedance: 10 GV, Common Mode Rejection Ratio: 115 dB at 60 Hz, Band-pass (10–1000 Hz)).

Infrared emitting diode (IRED) skin surface markers were affixed bilaterally to the lateral lower extremities. Triangular sets of IRED markers were secured to the thoracic spine (approximately at spinous process of thoracic vertebra 7) and pelvis (base of sacrum). Triangular marker sets were also attached to thigh, lower leg, and foot. Single IRED markers were placed on the lateral malleolus, lateral epicondyle, and greater trochanter of the femur and lateral aspect of the shoulder. After a standing calibration trial, digitization of 18 virtual points on predefined anatomical landmarks was completed, including right and left thoracic spine, right and left anterior superior and posterior superior iliac spines, Data acquisition

Three-dimensional lower extremity and trunk motion were recorded at 50 Hz using two optoelectronic motion analysis sensors (Optotrak 3020, Northern Digital Inc., Waterloo, ON, Canada). Three-dimensional ground reaction forces and moments were sampled from two force plates (AMTI, Advanced Mechanical Technology Incorporation, Newton, MA) at 500 Hz,

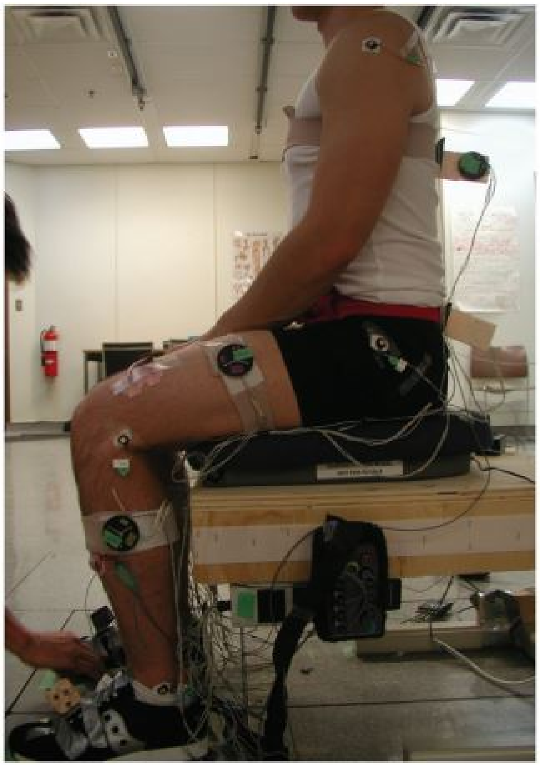

Figure 1. Left-sided sagittal plane view of experimental set-up. Triangular skin surface marker sets were placed in rigid circular discs and affixed to the foot, lower leg, thigh, pelvis, and thoracic spine as shown. Individual skin surface markers were placed on the shoulder, greater trochanter, lateral epicondyle, and lateral malleolus. Foot position was standardized on the two force plates for all trials. Knee angle was set to 90 _ prior to all trials and seat height was set to knee height as shown.

one placed under each foot. Both plates were aligned to global motion capture system coordinates. Participants were fully seated on the seat pan of the Power Seat device. Prior to each trial, both knees were positioned at 90 with a standard goniometer and feet were positioned facing forward and at approximately shoulder width apart. Participants remained stationary, arms across chest and trunk erect until the signal to stand was given. The rationale for not allowing participants to push off with their hands was to control the task to ensure consistency among participants in both groups between conditions. This allowed for meaningful comparisons (i.e. if we had different moments of force being produced among participants, between, groups, or between conditions, then no conclusion could be drawn regarding the alterations in lower limb kinematics, kinetics or EMG activity). The sit-to-stand protocol followed a semi-randomized design where all participants began the data collection completing five trials of regular sit-to-stand (ND). Participants were told to keep their feet placed on the floor, and look forward during all transfers. For the ND trials participants ascended to standing at self-selected pace when the signal to stand was given. Once standing, participants remained standing for the duration

of the trial recording. Following this series of trials (ND), the Power Seat trial set (PS) and a Seat Assist (UPE-1, Uplift Technologies Inc., Dartmouth, NS, Canada) set of five trials were randomly assigned using a coin flip.

For PS trials, participants pulled a lever to engage the seat when given the signal. Participants were instructed to keep their feet planted on the ground and remain on the seat until maximum elevation, and stand up at self-selected pace after the Power Seat stopped. Once standing, participants remained standing for the duration of the trial recording. Following the sit-to-stand trials, muscle activity was recorded during supine lying (resting bias). For EMG amplitude normalization and strength testing, participants completed at least one practice and two standardized three-second maximal voluntary isometric contraction (MVIC) trials. These exercises included knee extension at 45 of knee flexion from a supported sitting position. Ankle plantarflexion and dorsiflexion were measured in neutral ankle position from a long sitting position. Knee flexion was measured in prone-lying with the knee positioned at 55 of flexion. For each position, a gravity correction torque was determined. Torque output was collected using a Cybex Isokinetic dynamometer (Lumex, NY). Combined knee extension/hip flexion was also used to improve the likelihood of maximal levels of activity for rectus femoris [32]. For ES (trunk extension) and GM (leg extension), participants were positioned prone on the bed. Participant's trunk and lower limbs were strapped down to minimize the movement. Torques was not recorded for these trials. A 60-s rest period separated each contraction, and standardized verbal encouragement was given. All raw EMG and torque signals were acquired using a custom LabVIEW Version 9 (National Instruments Corporation, Austin, TX) data acquisition program; analog to digitally converted at 2000 Hz (16 bit, ± 5 V) and stored for later processing.

Data processing Raw EMG signals were processed through custom MatLab Ver 7.1 programs (The Mathworks Inc., Natick, MA). All signals (sit-to-stand and MVIC trials) were corrected for resting bias and converted to microvolts, full-wave rectified, low pass filtered (recursive Butterworth, 4th order, Fc-6 Hz). Maximum EMG amplitudes for each muscle during MVIC exercises were calculated using a 100 ms moving-average window (99 ms overlap) [30]. Maximum EMG amplitudes, regardless of MVIC exercise in which it occurred, were used for amplitude normalization of sit-to-stand EMG waveforms (% MVIC). Isometric torque values were corrected for the effect of gravity. Maximum torque for each exercise was identified using a 500 ms movingaverage window (0 ms overlap). The average of two trials was recorded as a muscle strength in Newton-meters and normalized to body mass (Nm/kg). Technical and local anatomical bone embedded coordinate systems for the trunk, pelvis, thigh, tibia, and foot were derived from IRED markers and digitized points. Joint angles were specified through Cardan/Euler rotations [33]. Trunk motion was described relative to the pelvis, where all other motions are described as the distal segment moving about the proximal segment (i.e. tibia relative to femur). Net external ankle, knee, and hip moments were calculated using an inverse dynamics model which combined ground reaction force and moment data, limb kinematics, limb anthropometrics, and inertial properties [34] and normalized to body mass (Nm/kg). The start of the sit-to-stand transfer was determined by visually identifying when the horizontal velocity of the trunk increased above baseline prior to reaching maximum horizontal velocity. An electric switch placed between the buttocks and the seat determined seat off. Sit-to-stand transfer termination was identified by the sample where the vertical velocity of the shoulder marker was zero after reaching maximum vertical velocity [35,36]. EMG, sagittal plane angles, and net external moments were time normalized to 100% from seat off to stand. All time normalizations were completed using a cubic spline interpolation. Ensemble averages were calculated from at least three sit-to-stand trials.

Table 1. Waveform variables used in the sit-to-stand transfer analysis.

Category	Variables
Sagittal plane angles	Peak (trunk, hip, knee, an ankle)
Sagittal plane net external moments	Peak (hip, knee, and ankle)
Electromyography	Peak and average muscle activity

Statistical analysis

LEFS scores were tabulated. Student t-tests were used to test for significant differences in age, mass, height, BMI, and LEFS scores. For muscle strength, a two-factor mixed model analysis of variance model (ANOVA) tested for significant group (between) and leg (within) main effects and interactions (alpha ¼ 0.05). Table 1 illustrates the response variables extracted from the joint angle, moment of force, and EMG waveforms used for analysis. Normality and equal variance of the response variables were determined from Kolmogorov–Smirnov and Levene's tests, respectively. All data with unequal variances or non-normal distributions were transformed using either a natural logarithm or an inverse hyperbolic sine function.

For the sit-to-stand joint angle, moment of force, and EMG waveform statistical analysis, group (younger and older adult), leg (right and left), device (no device (ND) and Power Seat (PS)), differences were tested using a mixed model analysis of variance that accounted for between and within group main effects and interactions (alpha ¼ 0.05). Post-hoc testing was employed for determining pair-wise significant findings using Bonferonni adjusted alpha levels. Statistical procedures were completed in Minitab Version16 (Minitab Inc., State College, PA).

Results

Twenty healthy participants were recruited. Group demographics and anthropometrics are shown in Table 2. Significant differences were found between the young and older adult groups for age and LEFS score (p50.05).

Muscle strength is shown in Table 3. No significant differences were found between right and left legs or between groups for knee extension and ankle dorsiflexion. Older adults had lower strength for the knee flexors (p ¼ 0.023) (hamstrings) and ankle plantarflexors (p ¼ 0.045) (gastrocnemius and soleus) when compared with younger adults.

The sit-to-stand transfer

For the PS and ND conditions, participants stood up from a stationary seat pan. The PS seat pan back elevated approximately 15 cm and moved forward approximately 9 cm during the PS condition. The Seat Assist device moved to the same position as the PS, however, operated on pneumatic pressure set to participant mass (i.e. when the participant initiated the sit-to-stand, the seat assist released and elevated). In every instance, participants stood up faster than the Seat Assist elevated. The seat off trigger system was affected by how fast participants stood up in addition to contact with the seat pan. Given data analysis occurred between seat-off and standing, this period had to be defined and for the

Seat Assist condition, the seat off indicator was influenced differently compared with PA and ND conditions. Therefore, we did not include the Seat Assist condition in the analysis.

Table 2. Participant demographics [mean (SD)].

	Young adults	Older adults
N	10	10
Percent Female	50%	50%
Age (years) (range)*	25	69
Mass (kg)	77 (13)	75 (15)
Height (m)	1.77 (0.08)	1.70 (0.10)
BMI (kg/m^2)	24.3 (2.6)	25.8 (4.0)
WHR	0.79 (0.07)	0.89 (0.09)
LEFS (/80)*	79.9 (0.3)	73.1 (4.8)

BMI, Body Mass Index; WHR, waist to hip ratio; LEFS, lower extremity functional scale.
*Significant group difference ($p < 0.05$).

Table 3. Isometric muscle strength (Nm) [mean (SD)].

	Young adults		Older adults	
Muscle group	Right	Left	Right	Left
Knee extensors (KE45)	148.1 (31.8)	160.5 (37.5)	129.0 (30.9)	123.5 (35.6)
Knee flexors (PKF55)*	82.5 (20.2)	84.8 (19.8)	63.0 (20.1)	62.2 (19.9)
Ankle plantarflexors*	100.1 (25.8)	107.3 (27.7)	77.0 (27.6)	79.1 (31.5)
Ankle dorsiflexors	39.8 (7.5)	41.3 (6.8)	36.4 (13.6)	37.2 (13.0)

*Significant group difference ($p < 0.05$).

Table 4. Peak joint angles and moments of force during the sit-to-stand transfer [mean (SD)].

Group	Condition	Hip	Knee	Ankle	Trunk (spine)
Sagittal plane joint angles (degrees)[a]					
Young adults	ND	81 (12)	78 (7)	−15 (4)	28 (9)
	PS	56 (8)	63 (7)	−19 (4)	15 (11)
Older adults	ND	91 (9)	81 (14)	−11 (4)	25 (8)
	PS	67 (7)	61 (14)	−14 (3)	10 (8)
Sagittal plane joint moments (Nm/kg)[b]					
Young adults	ND	0.90 (0.19)	0.62 (0.11)	0.14 (0.05)	
	PS	0.58 (0.20)	0.64 (0.12)	0.02 (0.06)	
Older adults	ND	0.90 (0.17)	0.56 (0.13)	0.12 (0.05)	
	PS	0.60 (0.14)	0.58 (0.14)	0.02 (0.06)	

ND, no device; PS, power seat.
[a]Hip, flexion (+); knee, flexion (+); ankle, dorsiflexion (−); trunk, flexion (+).
[b]Hip, flexion (+); knee, flexion (+); ankle, plantarflexion (+).

Mean (SD) total sit-to-stand duration for ND and PS in younger adults was 1.8 s (0.3 s) and 1.7 s (0.2 s), respectively, and 2.1 s (0.5 s) and 2.1 s (0.5 s) in the older adults, respectively. The ND and PS seat off to stand duration in younger adults was 1.1 s (0.1 s) and 1.0 s (0.1 s) and in older adults, 1.2 s (0.3 s) and 1.3 s (0.4 s), respectively.

Joint angles No significant leg main effects or interactions pertaining to leg were found (p40.05). Peak sagittal plane knee angles are shown in Table 4. Older adults performed the sit-to-stand transfer with approximately 10 greater peak hip flexion angle and approximately 5 less ankle dorsiflexion, which were significantly different from younger adults (p50.05). Peak knee flexion and trunk flexion angles were not different between groups. Peak hip and trunk flexion angles were significantly lower for the PS compared with ND (p50.05). Peak knee flexion angle was reduced by approximately 18% for the PS compared with ND (p50.05). Peak dorsiflexion was approximately 5% greater for the PS compared with ND, which was significant (p50.05).

Joint moments

A significant leg main effect was found for peak net external knee flexion and plantarflexion moments, where the right leg had greater moments than the left (p50.05). Lower peak net external hip flexion moments were shown for the right leg but this was not significant (p ¼ 0.08). No leg by device or leg by group interactions were found (p40.05). No group differences in peak net external flexion moments at the hip, knee, or peak plantarflexion moments at the ankle were found (p40.05). Table 4 provides the peak net external moment values for each device and between young and older adults. Peak net external hip flexion moments were significantly greater for ND, compared with PS (p50.05). Compared with ND, the peak hip flexion moments were reduced by approximately 35%. No differences in net external knee flexion moments were found between conditions. Compared with ND, the peak plantarflexion moments were reduced by approximately 85% with the PS (p50.05).

Muscle activation

Unequal variance and non-normality existed in EMG data. All data were transformed prior to statistical analysis. Peak and average muscle activations are found in Table 5. For peak activity of each muscle, no differences were found between legs for the younger and older adults (p40.05). Older adults had greater peak and average muscle activation amplitudes for the vastus medialis, lateral hamstrings, gluteus maximus, and the erector spinae muscles during the sit-to-stand transfer for both ND and PS conditions (p50.05). Greater average vastus lateralis activity was found in older adults for both transfer conditions (p50.05) whereas the peak activity did not differ (p40.05). VL waveforms are found in Figure 2. The power seat reduced the peak and average muscle activity for each muscle analyzed (p50.05).

Discussion

The sit-to-stand transfer is one of the most functionally demanding activities that individuals perform daily. Adaptive seating options exist, however, rarely are they tested for their mechanical effectiveness for assisting the sit-to-stand transfer prior to being offered to health care professionals and patients. To provide a foundation for this testing, the present study sought to test the hypothesis that trunk, and lower limb joint angles, moments of force and muscle activity, as a measure of mechanical demand, will be reduced with the assisted sit-to-stand and these differences will be similar between healthy younger and older adults. Mechanical demands were lower in younger and older healthy adults with the use of the Power seat, an adaptive seating system that provided mechanical assistance for the sit-to-stand transfer. This reduction was not, however, universally found among all variables suggesting that certain mechanical variables should be considered prior to prescription and use. Younger and older adults differed in age, LEFS scores, and knee flexor and ankle planterflexor strength. The knee extensors also generated approximately 25% less torque; however, these strength differences had little impact on the time taken to complete the-sit to-stand transfer. Total sit-to-stand time for younger and older adults was lower than previously reported [36]; a result thought to reflect differing time signature identification methods yet little difference (0.3 s) was found between the age groups. Ikeda et al. [15]

found that younger and older healthy adults completed the sit-to-stand transfer with similar timing, sagittal plane joint range of motion (ankle, knee, hip, and trunk to pelvis) and moments (knee and hip). In the current study, older adults had greater peak hip flexion and lower peak ankle dorsiflexion angles in both conditions. This suggests that a strategy to move the trunk over the feet may have been used during the transfer. In addition, older adults worked at a greater percentage of maximal muscle effort for quadriceps, lateral hamstrings, gluteus maximus, and the erector spinae; key muscles involved in sit-to-stand. While muscle activation onsets [18,37,38] and duration [28] have been reported for the sit-to-stand, amplitude levels (%MVIC), to our knowledge, have not been compared between healthy older and younger adults. The current results suggest older adults transfer with different hip and ankle kinematics and with greater muscle effort than younger adults; however, these group differences were not dependent on the use of the lifting seat device.

Table 5. Peak and average muscle activation (%MVIC) [mean (SD)].

	Young adults		Older adults	
Muscle	ND	PS	ND	PS
Peak activity (% MVIC)				
VL	59.4 (26.2)	36.4 (11.6)	69.3 (20.4)	50.5 (17.7)
VM	52.3 (13.7)	36.2 (10.8)	78.7 (24.7)	55.7 (23.7)
ES	36.4 (9.8)	19.0 (4.3)	48.9 (15.4)	31.5 (13.4)
GM	19.3 (6.9)	15.3 (7.1)	33.2 (13.1)	23.7 (7.9)
LH	12.0 (4.7)	7.7 (4.0)	24.3 (7.4)	16.1 (5.5)
TA	51.1 (20.8)	23.6 (15.0)	57.8 (20.5)	37.3 (15.5)
Average activity (% MVIC)				
VL	27.0 (6.9)	21.0 (5.4)	40.0 (12.5)	30.0 (7.6)
VM	24.0 (5.0)	19.2 (5.0)	42.1 (13.5)	29.6 (8.2)
ES	17.3 (4.5)	11.6 (4.0)	28.1 (10.7)	21.6 (11.6)
GM	13.0 (5.7)	11.1 (5.7)	21.8 (7.0)	17.0 (5.1)
LH	7.1 (2.5)	5.1 (2.6)	15.3 (4.9)	10.1 (3.3)
TA	13.2 (5.0)	5.9 (3.9)	16.3 (9.1)	9.2 (5.3)

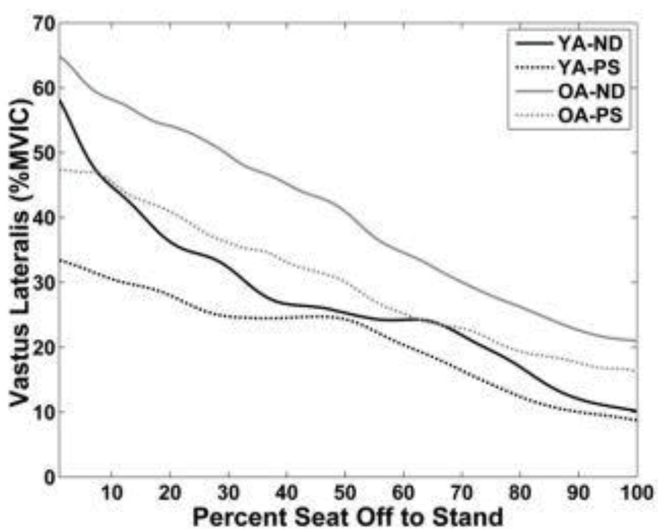

Figure 2. Group ensemble averaged vastus lateralis EMG for each seating condition (right and left legs collapsed). Percent MVIC is on the y-axis and percent seat off to stand on the x-axis. Dashed lines represent the power seat (PS) condition. Solid lines represent the no device (ND) condition. Dark lines represent young adult (YA) group and light lines represent the older adult (OA) group.

Previous studies focused on understanding the biomechanical advantages of adaptive seating systems that offer mechanical assistance using spring-based designs that aim to offer a mechanical boost during the transfer. In the current investigation, participants were instructed to stand up once the seat pan stopped moving. During elevation, the PS moved in the vertical direction 15 cm and in the anterior direction 9 cm. The PS was designed to have the seat pan hinge at two locations during elevation such that the back approximately one-third of the seat remains parallel to the floor – which differs from previously studied designs that only have an anterior hinge [20,23,25]. Compared with others [20,23], the PS did not provide a horizontal or vertical assistance thrust; a previous consideration for the application of these seats [20,23,25]. A time delay was found between PS stopping and when the participants stood up for every trial that was collected, indicating the absence of mechanical thrust to assist sit-to-stand.

In the present study, completing the PS transfer required less trunk, hip, and knee flexion range of motion, consistent with previous literature when high and low chair heights were tested [10,11,20]. The reduced range of motion requirements was accompanied by reduction in muscle effort suggesting lower demands on the musculoskeletal system during the PS transfer. In contrast, greater peak ankle dorsiflexion range of motion was required for both groups during the PS transfer. This occurred as a result of forward seat motion (9 cm) generated by the PS during ascent. Given the feet remained stationary, the participants' body moved forward about the ankle joint, causing greater dorsiflexion. As shown in Table 4, these dorsiflexion values exceed active range of motion values provided in the literature for adults of similar age [39] suggesting (1) functional range of motion may be different from active range of motion measured during a clinical exam; and (2) talocrural range of motion is an important consideration in the prescription of this device. An alternative would be to move feet farther forward during the transfer; however, a forward foot position has been shown to increase the knee and hip demands of accomplishing the sit-to-stand transfer [22]. Ankle dorsiflexion range of motion should be measured and without approximately 15–20 degrees, mechanical stress at proximal joints may result as a compensation for this limitation.

Sit-to-stand moments, considered surrogate measures for joint tissue stress, were also altered with PS use. Consistent with the reduced hip flexion range of motion found during the PS rise, sagittal plane peak net external hip moments were reduced approximately 35%. Reduced maximum hip moments have been previously found when rising from a higher seat [11,21,22]. The current findings suggest PS can provide benefit for individuals attempting to protect their hip, for instance with hip osteoarthritis or as temporary measure after total hip replacement. In contrast, sagittal plane peak net external knee moments did not differ between seat devices. Results in the literature are equivocal; however, in the current study, findings could not be explained by knee range of motion findings or knee muscle activation findings as a range of motion and quadriceps activation amplitudes both reduced with PS use. Given this reduction, a concomitant reduction in the sagittal plane knee moment was also expected. While a mechanism for this result was not clear, foot position and trunk position during the transfer can have implications for sagittal plane knee moments of force. While contrary to previous beliefs on the effects of this lifting seat device for individuals with knee impairments, these results suggest that mechanical knee joint stress may not be minimized with the use of the PS.

Biomechanical outcomes of this investigation suggest a lower overall mechanical demand is placed on the hip and trunk during the sit-to-stand transfer using the PS; however, there were no systematic decreases for the knee and ankle. The musculature also has implications for understanding mechanical demand as many individuals with lower extremity impairments present with weakness and muscular fatigue that limit sit-to-stand ability. Peak and average amplitudes of all muscles investigated were reduced when using the PS in both groups. The demands on the knee, hip, and the lumbar spine extensors are much less with PS use. Others have reported reduced quadriceps activity with sitto-stand from a high seat [21,28]. While significant reductions in the hip extensors were found in this study, the magnitude of change was less than found for the knee extensors. Arborelius et al. [21] found that medial hamstring activation did not change with increased seat height. Our findings do not support this result and could possibly be explained by our investigation of biceps femoris and different seat height levels and seating positions used between studies. ES assists the sit-to-stand transfer by extending the spine in collaboration with knee extension, and hip extension. In the current study, the ES reached a peak activity of approximately 36% MVIC and 49% MVIC for younger and older adults, respectively. To our knowledge, ES has not been investigated during the sit-to-stand transfer despite the demands on the trunk from a balance control perspective, owing to the dominant focus on the lower extremities in previous studies. These results suggest that this muscle group contributes significantly to this movement and should be considered in future work. Impairments that affect ES activation may have implications for the efficient movement of sit-to-stand. In addition, the PS reduced this activity, suggesting not only does this device reduce lower extremity demands but also the demands of the lumbar spine. This reduction in trunk flexion range and ES activity has implications for lifting seat device prescription for individuals with low back impairments. Given that overall activity was reduced in every muscle, the energy expenditure required per sit-to-stand transfer with the PS would theoretically be lower, reducing fatigue, and metabolic demands associated with the transfer task.

The findings of this comprehensive lower extremity and trunk biomechanical and neuromuscular sit-to-stand investigation were novel, providing both a foundation for further adaptive seating research and information for clinical prescription and device design. Interpretations still require the consideration of study limitations. Participants were provided with standardized instructions as per the guidelines accompanying the PS. All participants could rise from the chair unaided and, therefore, the arms were not used to (1) avoid the difficulty in interpreting lower extremity demands, which was the key question, and (2) to allow comparisons with previous sit-to-stand investigations where the use of arms is rarely permitted. Munro et al. [20] found that impulses obtained from arm chair rests were highly variable across each condition involving the ejector seat, reflecting a variety of compensatory techniques when performing the sit-to-stand. The arms role was different depending on the seat. During the high seat conditions, the arm use was thought to provide a stability role rather than a force augmentation role to assist the transfer [20]. This made the lower extremity biomechanics data difficult to interpret across conditions, given the varied use of the arms for support or force generation. For these reasons, arms were crossed at the chest during all sit-to-stand transfers to minimize confounding factors in this study.

Part of this investigation was to test the Seat Assist device, however, we realized that during the investigation, limitations arose that questioned our ability to include these tests in the statistical analysis. The Seat Assist works on pneumatics. As the participant begins to unweight the seat during the flexion momentum phase, the seat begins to rise slowly using the pneumatic pressure to assist with the sit-to-stand. This may be ideal for those who cannot get out of a chair, but for the healthy individuals in this study, they simply stood up and the seat pan came up behind them. In many cases, the participants were fully standing when the seat assist finished moving. It was difficult to have participants slow down and ride the Seat Assist upward, as it did not generate enough force to completely lift the participant. As a result, the participant would have to match the force of the Seat Assist, making the application unrealistic. Hence, we did not include the Seat Assist analysis and results in this manuscript.

Lifting seat devices are typically prescribed for individuals unable to transfer from sit-to-stand. Using these data, we are limited in drawing conclusions across all users of these devices. However, our results suggest that if individuals have at least 20 degrees of ankle dorsiflexion, benefit would be realized for those with hip injuries and disease, as well as those that require less low back muscle activation during sit-to-stand. Our understanding of how individuals without difficulty with sit-to-stand use these devices has been enhanced through this work. These data show that ankle and low back considerations are important for device prescription and that knee and hip joint mechanical demands are not systematically reduced with lifting seat use. This provides a foundation for future studies to better understand how individuals with inability to transfer from sit-to-stand utilize adaptive seating devices.

Conclusion

In conclusion, most biomechanical factors as well as lower extremity and trunk muscle activation levels during the sit-to-stand transfer task were altered with a mechanical device that raises the height of the seat pan to assist this transfer. Results partially confirmed our hypothesis in that knee, hip and trunk joint flexion angles, hip flexion, and ankle plantar flexion moments and all muscle activities were reduced with the assisted sit-to-stand and the effects were similar between healthy younger and older adults. Knee flexion moments were not altered with assisted sit-to-stand and greater ankle dorsiflexion was required. These results indicate that the raised seat did not have a systematic effect on lower extremity mechanics in healthy young and older adults where hip joint mechanics were most positively influenced with the use of the power seat. These results will be important to consider for future recommendations on the use of these products and research pertaining to the advancement of adaptive seating.

Declaration of interest

The authors acknowledge that there are no conflicts of interest pertaining to this manuscript. All authors contributed to study inception, data collection, analysis and writing. All authors have agreed upon the typeset. This work was supported by a National Sciences and Engineering Research Counsel of Canada (NSERC) Engage Grant (#EGP/418922-2011).

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