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Theoretical contribution of the upper extremities to reducing trunk extension following a laboratory-induced slip

Karen L. Troy, PhD, **Stephanie J. Donovan, MS**, and **Mark D. Grabiner, PhD**
Chicago, Illinois USA, Biomechanics Research Laboratories, Department of Kinesiology and Nutrition, University of Illinois at Chicago

Abstract

Slips are frequently the cause of fall-related injuries. Identifying modifiable biomechanical requirements for successful recovery is a key prerequisite to developing task-specific fall preventive training programs. The purpose of this study was to quantify the biomechanical role of the upper extremities during the initial phase of a slip resulting in trunk motion primarily in the sagittal plane. Two groups of adults were examined: adults over age 65 who fell and adults age 18–40 who avoided falling after slipping. We hypothesized that rapid shoulder flexion could significantly reduce trunk extension velocity, that adults who slipped would implement this as a fall avoidance strategy, and that younger adults who avoided falling would use this strategy more effectively than older adults who fell. The kinematics of the 12 younger adults and eight older adults were analyzed using a three-segment conservation of momentum model developed to represent the trunk, head and upper extremities. The model was used to estimate the possible contribution of the upper extremities to reducing trunk extension velocity. The model showed that upper extremity motion can significantly reduce trunk extension velocity. Although the upper extremities significantly reduced the trunk extension velocity of both young and older adults ($p<0.027$), the reduction found for the young adults, 13.6±11.4%, was significantly larger than that of the older adults (5.8±3.4%, $p=0.045$). Given the potential for trunk extension velocity to be reduced by rapid shoulder flexion, fall prevention interventions focused on slip-related falls may benefit from including upper extremity motion as an outcome whether through conventional or innovative strategies.

Introduction

Falls by older adults cause considerable mortality, morbidity and economic burden on the healthcare system in the United States. The risks of a fall-related fatality or hospitalization for an older adult are five and three times that of any other age group, respectively (Finkelstein et al., 2006). Up to 50% of fall-related injuries in older adults may arise from slipping (Courtney et al., 2001). Once a slip is initiated, avoiding a fall is a complex and time critical motor skill, the performance of which becomes increasingly difficult with age (Troy et al., 2008). It may be possible to develop effective interventions that reduce the incidence of slip-related falls.

Corresponding Author: Karen L. Troy, PhD, Department of Kinesiology and Nutrition (MC994), 1919 West Taylor St. Room 650, Chicago, IL 60612, email: klreed@uic.edu, Tel: 312.413.5495, Fax: 312.996.9296.

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However, doing so will require characterization of the modifiable biomechanical requirements of the recovery task.

Upper extremity elevation may be important to fall avoidance after a slip (Marigold et al., 2003; Tang and Woollacott, 1998). When a slip occurs while walking forward, although the body center of mass (COM) translates forward, the body rotates backwards. The head, arms and trunk (HAT) is 60 percent of body mass, suggesting that limiting trunk extension may help maintain the overall body COM within the base of support. Upper extremity motion could help limit trunk extension through two mechanisms. First, arm elevation moves the HAT COM anteriorly thereby increasing its distance from the posterior margin of the base of support. Second, rapid shoulder flexion may reduce trunk extension velocity.

This study quantified the role of the upper extremities during the initial phase of a slip, from heel-strike until the instant the recovery foot was placed on the ground. For comparison to previous studies that have induced slips using translating platforms (Tang and Woollacott, 1998) and rollers (Marigold et al., 2003), analysis was limited to slips resulting in trunk motion that was primarily in the sagittal plane. Older adults who fell following an induced slip were compared to younger adults who did not fall. Two questions were addressed: (1) To what extent can and do the upper extremities reduce trunk extension velocity after a slip? (2) Do the two groups use their upper extremities differently? We hypothesized that rapid shoulder flexion could significantly reduce trunk extension velocity and that adults who slipped would implement this as a fall avoidance strategy. We expected that younger adults who avoided falling would use their upper extremities to reduce trunk extension velocity to a greater extent than older adults who fell.

Methods and Materials

Subjects

The subjects whose slips were analyzed here were a subset from a larger study (Troy et al., 2008) consisting of 35 healthy young adults and 21 healthy older adults who provided written informed consent prior to participating in this institutionally reviewed and approved study. Older adults were independent community-dwellers screened by a physician for exclusion factors including neurological, musculoskeletal, and cardiovascular disorders. Of these subjects, the data of 30 young subjects who unambiguously recovered from the slip and 18 older subjects who unambiguously fell were eligible for inclusion in the present analysis. This subject pool was further narrowed by eliminating those whose trunk motion was not directed primarily backwards. Subjects who were included had a peak trunk extension velocity of >50 deg/s and a peak lateral trunk velocity of <50 deg/s during the period from slip onset to 133 ms after recovery foot touchdown. During this period no subject had yet engaged the safety harness and subjects who avoided falling generally had reversed the trunk extension velocity. The present analysis included12 young adults (5 males, mean±SD: 24±5 years, 168±11 cm, 66±11 kg) and 8 older adults (1 male, 71±4 years, 162±9 cm, 70±11 kg).

Experiment

Subjects traversed a 12.8 m walkway along which a Plexiglas sheet (1.22m x 2.44m x 0.63cm) was placed approximately mid-way. Subjects were informed that the Plexiglas was present to protect instrumentation embedded in the floor. Subjects first performed five normal walking trials during which there was no risk of slipping. After these trials subjects wore a safety harness with a dynamic rope attached at the shoulders to a ceiling-mounted track. The instrumented harness prevented contact with the ground if they fell and allowed verification of the extent to which the harness supported the subject. During the trials in which subjects wore the harness, they were aware that they might be slipped, but not when or where the slip would occur or how it would be induced. Subjects were instructed to walk as if they had a destination, and in the event of a slip, to regain their balance if possible.

The laboratory lights were dimmed to diminish reflections from the Plexiglas surface, but remained bright enough that subjects could easily see the environment. Immediately prior to the slipping trial, the subject was distracted while an investigator either applied a film of mineral oil to the Plexiglas surface, or sprayed the surface with a mist of water to activate a film of water-soluble lubricant that had been allowed to dry on the Plexiglas. The static coefficient of friction for both lubricants was measured as 0.19±0.13 depending on footwear. The dynamic coefficient of friction was an order of magnitude smaller. Most young subjects slipped on oil, while most older subjects slipped on lubricant. All subjects reported that they were unaware that the surface characteristics had changed prior to the slipping trial. Only one slip was attempted for each subject. For all trials, the motions of 23 passive reflective markers placed over anatomical landmarks (Kadaba et al., 1990) were collected at 60 Hz using an eight camera motion capture system (Motion Analysis, Santa Rosa CA).

Data Analysis—The slipping event was temporally bounded by the onset of the slip and recovery foot touchdown; both identified manually from motion capture data by a single investigator. Slip onset was defined as the instant of heel-strike on the slippery surface. Recovery was the instant at which the non-slipping foot contacted the floor. All subjects took a recovery step.

Sagittal plane kinematics extracted from the motion capture data included trunk extension angle and angular velocity (relative to vertical), and shoulder flexion angle and angular velocity (relative to the trunk; Orthotrak 6.2, Motion Analysis, Santa Rosa, CA). All shoulder data are referenced to either the slip-side or recovery-side limbs.

The anterior/posterior position of the HAT COM relative to the greater trochanter, X_{HAT}, was calculated from the segment masses (Winter, 1990) and positions at the instants of slip onset and recovery to determine whether subjects used their upper extremities to translate the HAT COM anteriorly. A positive X_{HAT} indicated that the HAT COM was anterior to the greater trochanter.

Theoretical Model

To quantify the contribution that rapid shoulder flexion could make towards reducing trunk extension, a three-segment sagittal plane model was developed based upon conservation of angular momentum. The three segments represented the head/neck/trunk and the combined right and left upper extremities. The model assumed that the only external forces acting on each segment were gravity and, at the distal end of the trunk, the reaction force and moment. It was also assumed that the linear velocity of the shoulder center of rotation remained constant and that angular momentum of the system about the shoulder center of rotation was conserved. To avoid gross violation of these assumptions, analysis was limited to the period between slip onset and recovery foot touchdown. The model was based on the following equation:

$$I_T \omega_T + I_a * (\omega_{ra} + \omega_{la}) = \text{constant} \tag{1}$$

Where I is the moment of inertia of the trunk (T) and arm (a) segments, and ω is the angular velocity. The segment moments of inertia were calculated about the shoulder center of rotation and segment angular velocities were referenced to the global reference frame.

The change in angular velocity of the trunk, $\Delta\omega_T$, is governed by the following equation:

$$I_T\,\Delta\omega_T + I_a * (\Delta\omega_{ra} + \Delta\omega_{la}) = 0 \quad (2)$$

We assumed an initial (theoretical) condition in which trunk extension velocity, ω_{T0} was not zero but that there was no shoulder flexion velocity. That is,

$$\omega_T(0) = \omega_{T0}$$

$$\omega_{ra}(0) = \omega_{la}(0) = 0$$

If the arms started at zero velocity and moved to their peak velocities, they would impart a maximum change to the angular momentum of the trunk. To calculate this, we assumed that for each segment:

$$\Delta\omega_{max} = \omega_{peak} - \omega(0)$$

where ω_{peak} is the measured peak angular velocity for a segment. Thus, the maximum possible change in trunk angular velocity imparted by moving the arms at their peak velocities would be: $\Delta\omega_{T\ max} = \omega_{Tpeak} - \omega_{T0}$, where ω_{T0} is calculated as:

$$\omega_{T0} = I_a / I_T * (\omega_{rapeak} + \omega_{lapeak}) + \omega_{Tpeak} \quad (3)$$

Similarly, at any given time point t, $\Delta\omega_T(t) = \omega_T(t) - \omega_T(0)$. Substituting this into equation (2) and solving for the theoretical $\omega_T(0)$ given the actual $\omega_T(t)$ at each time produced the dotted line in Figure 1 when the actual arm angular velocities ($\omega_{ra}(t)$ and $\omega_{la}(t)$) were used at each time point. The dashed line was produced when the maximum observed arm angular velocities (ω_{rapeak} and ω_{lapeak}) were used at each time point.

First, the model was used to estimate the shoulder flexion velocity required to reduce the angular momentum of the trunk by a specific proportion, *p*, given an initial ω_{T0}. This was accomplished by solving for the quantity ($\omega_{rapeak} + \omega_{lapeak}$) assuming that $\omega_{Tpeak} = (1-p) * \omega_{T0}$. Then the corresponding mean peak shoulder velocity was computed using this value and ω_{Tpeak}. Second, the model was used estimate the maximum possible contribution of upper extremity motion to reducing the trunk extension velocity for each subject during the slip trial, $\Delta\omega_{T\ max}$, and the reduction in trunk extension velocity at recovery, $\Delta\omega_T$(recovery). This was accomplished by using subject-specific moment of inertia values for the total arm and head-neck-trunk segments (Winter, 1990) to calculate a theoretical $\Delta\omega_{T\ max}$ given a subject's actual ω_{rapeak}, ω_{lapeak}, and ω_{Tpeak} or a theoretical $\Delta\omega_T$(recovery) given a subject's actual ω_{ra}(recovery), ω_{la}(recovery), and ω_T(recovery), (Figure 1).

The hypotheses that rapid shoulder flexion could reduce trunk extension velocity and that adults would implement this strategy were tested using a one-sample t-test to compare the theoretical $\Delta\omega_{T\ max}$ of each group to a value of zero. One-sample t-tests (versus zero) were also used to test whether each group flexed their shoulders (to shift the HAT COM forward) and to test whether X_{HAT} changed during the slip. The hypothesis that young non-fallers would reduce trunk extension velocity more than old fallers was tested by a between-groups comparison of $\Delta\omega_{T\ max}$ using an independent t-test. To determine other between-group differences, kinematic variables at slip onset and recovery, and peak values of each variable were compared using independent t-tests.

Results

Although all young adults recovered and all older subjects fell after slipping, the between-group differences for the event duration (317±54 ms for young, 285±57 ms for older subjects) and for most kinematics were not significant ($p>0.05$; Table 1 and Figure 2). Young adults had larger trunk extension angles at slip onset, recovery, and at peak compared to older subjects (Table 1). Compared to the young adults, X_{HAT} was more anteriorly positioned in older adults at slip onset (11.6±9.5 cm versus 0.1±15.4 cm, $p=0.05$), but not at recovery (6.2±8.6 cm versus 0.7±13.6 cm, $p>0.05$). During the slip the HAT center of mass translated significantly backwards relative to the hips in older adults ($\Delta X_{HAT}=-5.4$, $p=0.016$) but not in younger adults ($\Delta X_{HAT}=0.53$, $p=0.936$) Peak recovery-side shoulder flexion angle was 10.9±14.2 and 10.4 ±13.8 degrees for younger and older subjects, respectively. These values were different from zero in younger ($p=0.027$), but not older ($p=0.075$) subjects. Slip-side peak shoulder flexion angle was not different from zero in either group ($p>0.455$).

The model indicated that young non-fallers were not more effective than old fallers at using their upper extremities to reduce trunk extension velocity. However, one older subject, identified as a statistical outlier, had a reduction in peak trunk extension velocity of 42.5%. This reduction of peak trunk extension velocity was over four times that of the next nearest subject. In the absence of this subject, the between-group difference was significant (Table 1). The model indicated that rapid shoulder flexion could decrease trunk extension velocity following a slip. $\Delta\omega_{Tmax}$ was significantly different from zero for both groups ($p<0.027$). Non-parametric tests produced similar between-group results. According to the model, the shoulder flexion velocity required to decrease the trunk extension velocity by a given percentage increased linearly as a function of trunk extension velocity (Figure 3). Only one older subject, the outlier, achieved a greater-than-9.5% reduction in trunk extension velocity. In contrast, seven younger adults achieved this magnitude of reduction (Figure 3).

Discussion

We quantified the biomechanical role of the upper extremities during the initial phase of a slip in the sagittal plane and found that upper extremity motion can contribute to fall avoidance. This occurs by increasing the distance between the HAT COM and the posterior margin of the base of support and by reducing trunk extension velocity. The results supported all three hypotheses and indicated that upper extremity motion can contribute to the recovery effort after a sagittal plane slip, that both young and older adults implemented this strategy, and that the younger adults who avoided falling were more than twice as effective at reducing trunk extension velocity than the older adults who fell. Similar to a previous report (Marigold et al., 2003; Tang and Woollacott, 1998), subjects elevated the upper extremities during the slip. However, only the recovery-side peak shoulder flexion of younger adults was significantly different from zero. This suggests that during the initial phase of a slip, anterior translation of the HAT COM via upper extremity motion may not be central to the problem of avoiding a fall.

Younger adults may be more effective at reducing trunk extension velocity because of the ability to rapidly flex the shoulders. The largest single peak shoulder flexion velocity achieved by a young subject was 599 deg/s. The largest average of the peak velocities of the slip and recovery-side shoulders was 394 deg/s. In contrast, the largest peak shoulder flexion velocity for an older adult was 356 deg/s and the largest average of the peak velocities of the slip and recovery-side shoulders in older adults was 237 deg/s.

The findings point to a potentially modifiable difference in performance capabilities that distinguished the younger and older adults. Differences in muscle strength and power, reaction and response times, or other factors may underlie the more effective use of shoulder flexion moments by the young adults to reduce trunk extension velocity following a slip. If so, these may represent viable areas for clinical intervention as they can be improved.

Our model supports recent emphasis on the importance of the upper extremities in recovery of balance after a slip (Marigold et al., 2003; Tang and Woollacott, 1998). Despite the relatively small mass of the arms compared to the trunk, their rapid elevation may contribute to reducing trunk extension velocity induced by a slip. The young adults in our study benefitted to a greater extent from this strategy than the older adults with one exception, who was able to reduce trunk extension velocity by over 42% (Figure 3). Overall, though, the model and data suggest that most of the older adults we slipped were less effective at implementing this strategy.

Although all of the younger adults avoided falling and all the older adults fell, most trunk and shoulder kinematics during the analysis period appeared similar for the two groups. This may reflect the use of rapid trunk extension as a criterion by which subjects were selected for inclusion into this study. It is also possible that subjects modulated their upper extremity responses depending on the magnitude of the disturbance (Marigold et al., 2003). If so, our present data set would not distinguish an intentional (modulated) lack of response in young non-fallers from a potential inability to respond by older fallers.

Trunk extension angle was consistently larger in young non-fallers than in older fallers. This could reflect the generally greater flexibility of younger subjects compared to older adults. Larger hip joint range of motion would allow a longer backwards recovery step despite the kinematic limitations imposed on hip extension by a larger trunk extension. Younger adults may also simply “allow” themselves to have larger trunk extension angles during the slip. Indeed, young adults tend to be more willing to move their centers of mass close to the edge of their base of support before initiating a step in response to a disturbance (Patton et al., 2006).

There were a number of sources of potential error in our analyses. For example, because the moments of inertia for the trunk and the arm segments used in the model were based on anthropometric data, minor discrepancies between actual segment values and published values may be present. The model assumed the elbows remained extended and near-simultaneous peak left and right shoulder flexion velocity, the latter of which did not generally occur. Thus, the model estimates may be closer to the maximum possible contribution of the upper extremities than to the actual contribution. The model also assumed conservation of angular momentum about the shoulder center of rotation, constant linear velocity of the shoulder center of rotation, that each segment experienced gravitational force and that a reaction force and moment existed at the distal end of the trunk segment. Limiting the analysis to the time period between heel strike and recovery foot touchdown helped to minimize any violation of these assumptions. Although one would expect upper extremity motion to contribute to trunk kinematics beyond recovery foot touchdown, the moment created by the recovery foot ground reaction force would cause our model to no longer be valid.

In theory, shoulder flexion can make the most meaningful and beneficial contribution to slip recovery if it occurs rapidly and only up to where the arms are parallel to the ground. Shoulder flexion beyond parallel to the ground will decrease the anterior translation of the HAT from a maxima (relative to the hips) and may add to the trunk extension velocity. During the analysis period none of the subjects approached this degree of shoulder flexion; the peak shoulder flexion angle for any subject was 43 degrees. However, the upper extremities of both young and older subjects continue to move after the recovery foot contact, which may be important later in the recovery phase. Qualitative observations indicated rapid and symmetric arm elevation to a nearly horizontal position shortly after recovery by many subjects. The associated forward linear momentum combined with the ground reaction from the recovery limb, may have aided the recovery effort.

Clearly, factors other than upper extremity use contributed to the observed differences between the slip outcomes of the young and older adults. Only one older adult used the upper extremities to reduce trunk extension velocity by more than 10%. However, half the younger adults, all of whom recovered, had reductions of 10% or less due to upper extremity use. Reduction of trunk extension velocity by the upper extremities does not appear to be a determinant of the slip outcome. Nevertheless, because rapid shoulder flexion can significantly reduce trunk extension velocity, under some circumstances it can plausibly serve as a contributor to the recovery effort.

In conclusion, young adults who avoided falling used upper extremity motion to reduce trunk extension velocity after a slip. Most older adults who fell appeared less able to do so. A model showed that rapid shoulder flexion can reduce trunk extension velocity, and that the shoulder flexion need not be symmetric to accomplish this. Diminished neuromuscular performance may explain why the older adults did not reduce trunk extension velocity to the extent of young non-fallers. In aggregate, the results suggest that fall prevention interventions focused on slip-related falls may benefit from including upper extremity motion as an outcome.

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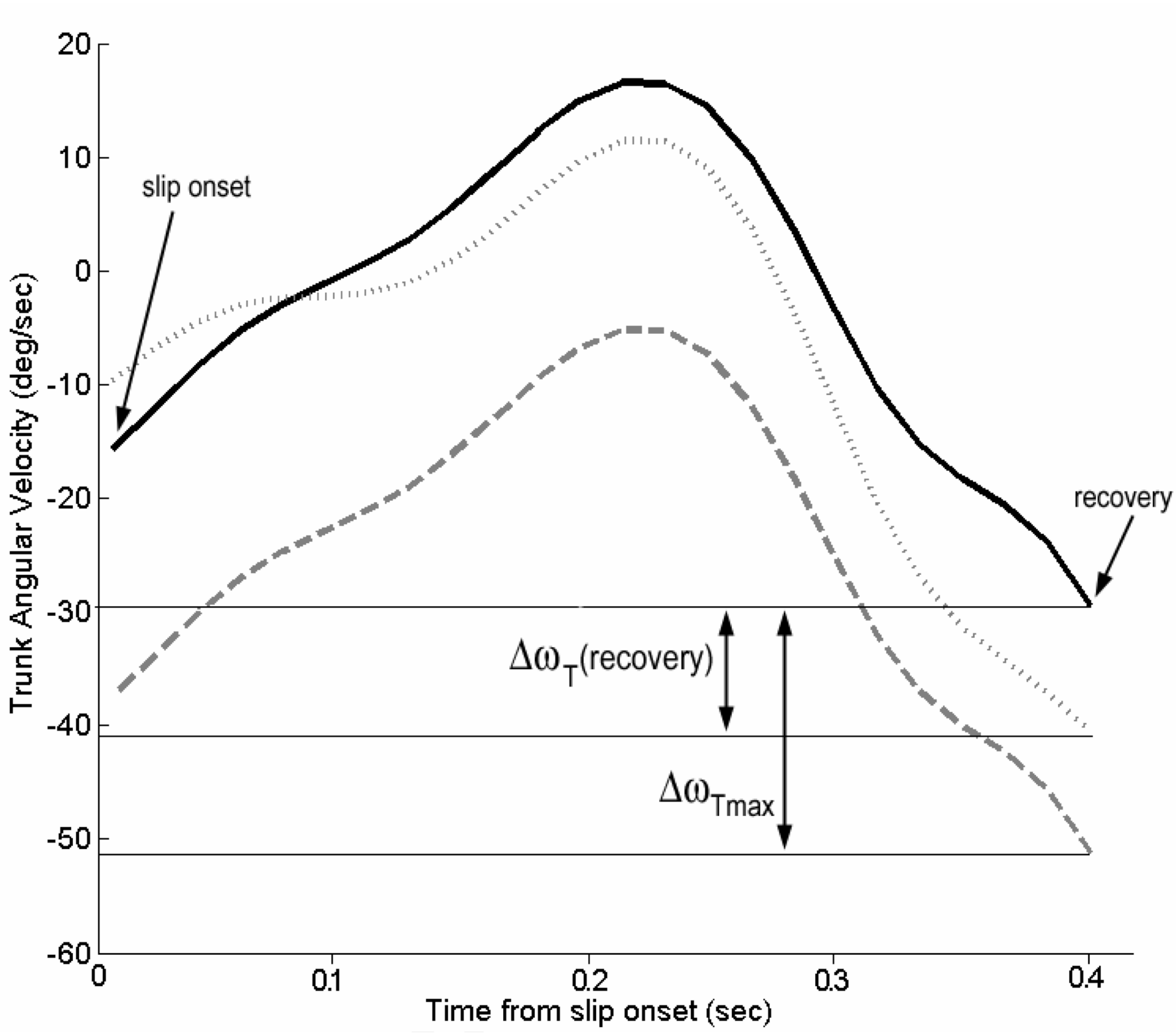

Figure 1.

The solid black line shows the actual trunk extension velocity (ω_T) of one older adult from slip onset to the instant of recovery (negative values indicate trunk extension). The dotted gray line shows the model-derived trunk extension velocity using the actual shoulder flexion velocities at each time point. The difference between the model-derived and the experimentally-derived data at the instant of recovery yields the variable $\Delta\omega_T$(recovery). The dashed gray line shows the model-derived trunk extension velocity calculated by using the maximum shoulder flexion velocity at each time point. $\Delta\omega_{Tmax}$ is the maximum possible amount that the upper extremities could reduce trunk extension and is calculated at the instant of peak trunk extension, which in this case is the final data point.

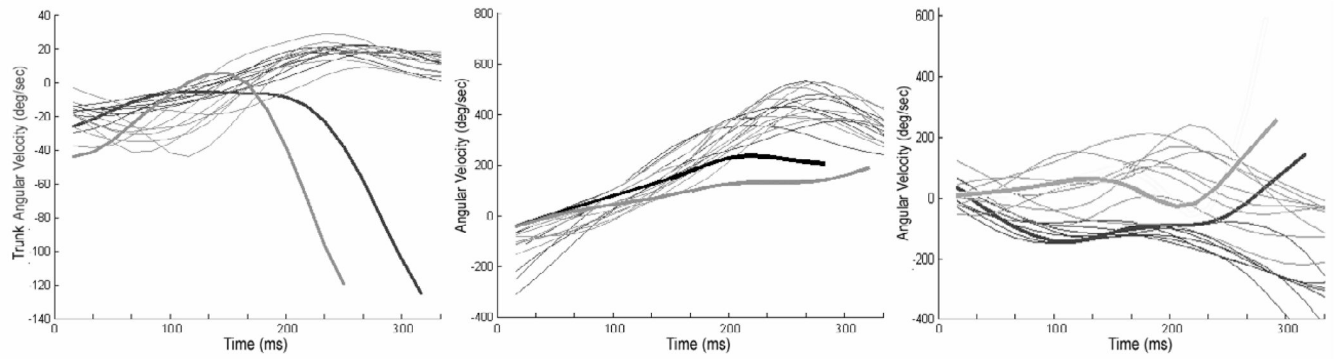

Figure 2.

Typical older (black) and younger (gray) adult angular velocity profiles for the trunk (left), slip-side shoulder (middle) and recovery-side shoulder (right). In all plots a negative angular velocity indicates extension and a positive velocity indicates flexion. Thin lines represent the first 333 ms after heel-strike during a normal walking cycle. The thick lines represent the velocity profiles from slip onset to recovery for the two subjects. Qualitatively, velocity profiles were similarly shaped for the older and younger adults during normal walking and slipping trials. In general, the arms and trunk did not deviate from their normal walking velocity profiles until 170 ms following the slip.

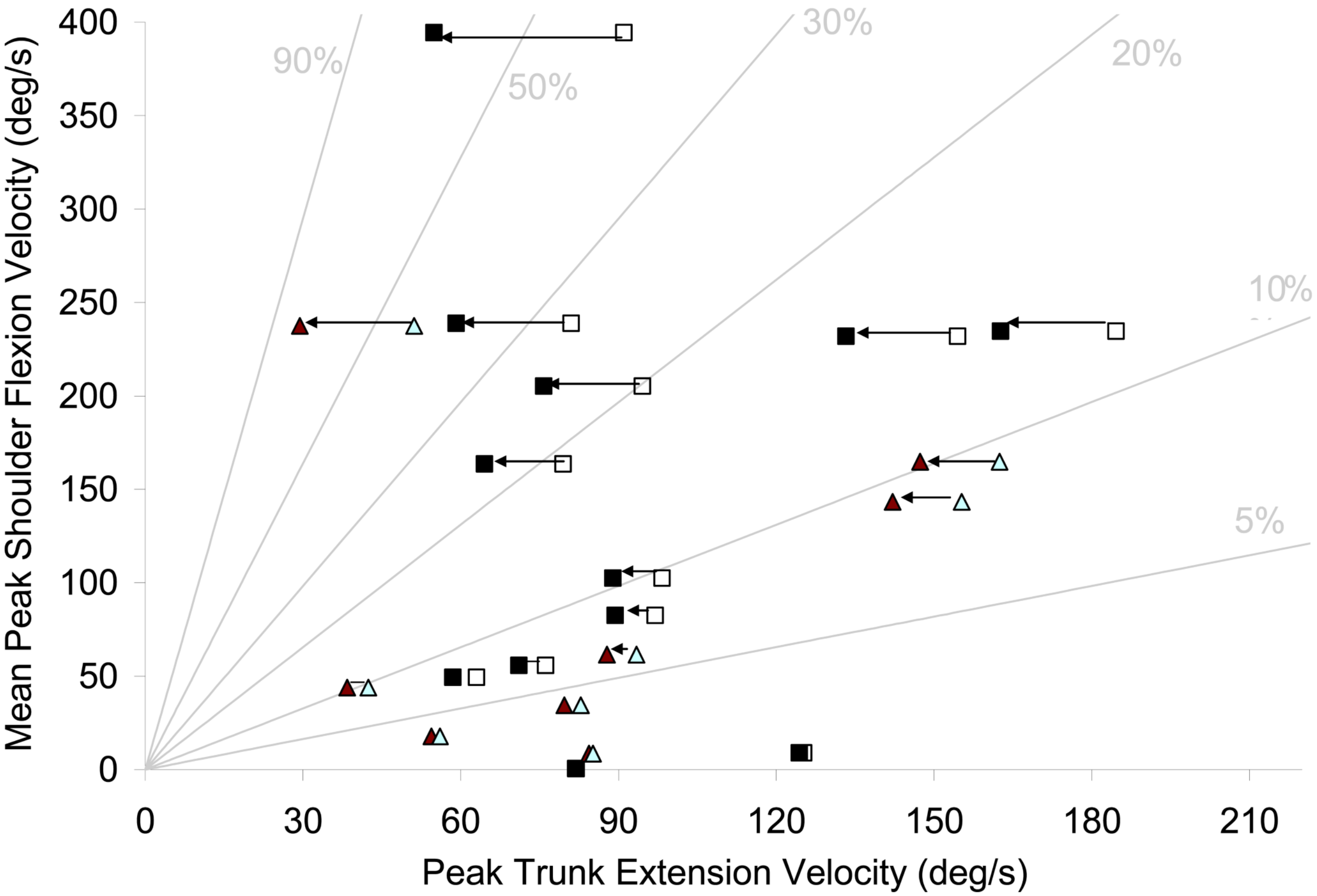

Figure 3.

Average of the peak slip and recovery shoulder flexion velocities versus trunk extension velocity. The gray contour lines correspond to the indicated percentage reduction of trunk extension velocity due to arm movement. Each subject is represented by two data points. The closed symbol represents the actual measured peak trunk extension velocity (ω_{Tpeak}). The corresponding open symbol, to the right, represents the model-derived ω_{T0} for that subject. The arrow between the data points reveals the magnitude of the the reduction of trunk extension velocity from ω_{T0} to ω_{Tpeak}. by the use of the upper extremities. Young subjects, who avoided falling, are represented with square points and older subjects, all of whom fell, are represented with triangle points.

Table 1

Kinematic and model-derived data for the 12 younger and 8 older subjects at the instants of slip onset and recovery, and the peak values. The p-values are from independent t-tests comparing groups. The last column indicates the total number of subjects in each group for whom the peak value of the variable occurred at the instant of recovery.

	Young	SD	Old	SD	p (young vs. old)	N (young, old) whose peak occurred at recovery
At slip onset						
Slip-side shoulder flexion angle (deg)	−16.8	16.3	−11.3	9.9	0.405	
Rec-side shoulder flexion angle (deg)	7.5	11.2	5.9	8.1	0.728	
Slip-side shoulder flexion velocity (deg/s)	−4.0	36.7	−28.7	34.0	0.147	
Rec-side shoulder flexion velocity (deg/s)	−34.8	24.4	−33.0	19.7	0.867	
Trunk extension (deg)	4.7	6.0	−1.7	3.2	0.014*	
Trunk extension velocity (deg/s)	16.3	13.4	22.6	13.2	0.311	
Peak values						
Slip-side shoulder flexion velocity (deg/s)	164.8	166.7	95.4	54.8	0.434	(0, 0)
Rec-side shoulder flexion velocity (deg/s)	129.8	122.5	82.6	144.2	0.564	(9, 3)
Trunk extension (deg)	11.9	7.2	4.5	5.8	0.025*	(12, 6)
Trunk extension velocity (deg/s)	88.7	34.1	82.9	43.7	0.744	(4, 6)
ωTpeak (deg/s)	95.0	34.3	86.7	44.4	0.645	
ΔωTmax (deg/s)	13.5	10.8	8.1	7.6	0.239	
ΔωTmax (%)	13.6	11.4	10.4	13.3	0.592**	(2, 2)
At recovery						
Slip-side shoulder flexion angle (deg)	2.1	25.4	−0.8	8.5	0.713	(8, 7)
Rec-side shoulder flexion angle (deg)	2.7	17.5	0.4	20.4	0.794	(4, 2)
Slip-side shoulder flexion velocity (deg/s)	75.9	147.8	40.9	186.6	0.751	
Rec-side shoulder flexion velocity (deg/s)	87.4	164.0	57.0	89.2	0.563	
Trunk extension (deg)	11.9	7.2	4.4	5.9	0.024*	
Trunk extension velocity (deg/s)	77.0	41.9	82.2	44.0	0.793	
ωT(recovery) (deg/s)	80.4	41.9	84.3	45.6	0.849	
ΔωT(recovery) (%)	12.4	27.2	5.3	10.8	0.876	

	Young	**SD**	**Old**	**SD**	**p (young vs. old)**	**N (young, old) whose peak occurred at recovery**

[*] difference between younger and older subjects is significant ($p<0.05$)

[**] when one older subject statistical outlier is removed from the data set, this value becomes $p=0.045$ and the theoretical reduction in trunk extension velocity, expressed as a percentage, decreases to 5.8% (SD: 3.4)

ωT0 is the theoretical peak trunk extension velocity in the absence of arm motion

$\Delta\omega$Tmax = ωT0 - [peak trunk extension velocity]

ωT(recovery) is the theoretical recovery trunk extension velocity in the absence of arm motion

$\Delta\omega$T(recovery) = ωT(recovery) - [recovery trunk extension velocity]