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Age-Related Changes in Lower Trunk Coordination and Energy Transfer During Gait

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McGibbon, Chris A. and David E. Krebs. Age-related changes in lower trunk coordination and energy transfer during gait. *J Neurophysiol* 85: 1923–1931, 2001. The effects of aging on lower trunk (*trunk–low-back joint–pelvis*) coordination and energy transfer during locomotion has received little attention; consequently, there are scant biomechanical data available for comparison with patient populations whose upper body movements may be impaired by orthopaedic or neurologic disorders. To address this problem, we analyzed gait data from a cross-sectional sample of healthy adults ($n = 93$) between 20 and 90 yr old ($n = 44$ elderly, >50 yr old; $n = 49$ young, <50 yr old). Gait characteristics of elders were mostly typical: gait speed of elders (1.13 ± 0.20 m/s) was significantly ($P = 0.007$) lower than gait speed of young subjects (1.20 ± 0.18 m/s). Although elders had less low-back (trunk relative to pelvis) range of motion (ROM; $P = 0.013$) during gait than young subjects, no age-related differences were detected in absolute trunk and pelvis ROM or peak pitch angles during gait. Despite similar upper body postures, there was a strong association between age and pelvis-trunk angular velocity phase angle ($r = 0.48$, $P < 0.001$) with zero phase occurring at approximately 55 yr of age; young subjects lead with the pelvis while elderly subjects lead with the trunk. Age related changes in gait speed and low-back ROM were unable to explain the above findings. The trunk-leading strategy used by elders resulted in a sense reversal of the low-back joint power curve and increased ($P = 0.013$) the mechanical energy expenditure required for eccentric control of the lower trunk musculature during stance phase of gait. These data suggest an age-related change in the control of lower trunk movements during gait that preserves upper body posture and walking speed but requires a leading trunk and higher mechanical energy demands of lower trunk musculature—two factors that may reduce the ability to recover from dynamic instabilities. The behavioral and motor control aspects of these findings may be important for understanding locomotor impairment compensations in aging humans and in quantifying falls risk.

INTRODUCTION

Locomotion in bipedal primates, humans in particular, is unique in the sense that control of the upright trunk is especially critical for maintaining control of balance (Krebs et al. 1992; MacKinnon and Winter 1993; Nashner and McCollum 1985; Winter 1987). The effects of natural aging may impact locomotor function due to age-related degeneration of the musculo-skeletal system (Frontera et al. 1991; Grimby 1995; Tinetti and Ginter 1988; Wolfson et al. 1995), sensory feedback systems (Allum et al. 1997; Peterka et al. 1990; Stelmach and Worringham 1985) and the brain's motor control structures (Rogers and Bloom 1985). Because the upper body constitutes two-third body weight, subtle uncoordinated movements of the upper body may affect the ability to recover from dynamic instabilities.

Studies of coordination in human walking have focused primarily on the lower limbs (Bianchi et al. 1998; Grasso et al. 1998, 2000; Grillner 1981) of young, healthy persons. Few studies have examined the upper body during gait and have focused primarily on trunk kinematics in the young and elderly (Krebs et al. 1992; Murray 1967; Opila-Correia 1990; Thorstensson et al. 1984) or pelvis and trunk kinematics in young adults (Stokes et al. 1989; van Emmerik and Wagenaar 1996). It is unknown what changes in pelvis-trunk (lower trunk) interactions, if any, occur with aging; consequently, there are scant biomechanical data available for comparison with patient populations whose upper body movements may be impaired by orthopaedic or neurologic disorders. We took as our primary hypothesis that aging in healthy (asymptomatic) adults affects the kinematic coordination of the lower trunk during gait, and we tested whether such alterations are modulated by age-related changes in kinematic gait parameters, such as walking speed and range of movement (Himann et al. 1988; Larish et al. 1988; Oberg et al. 1993; Ostrosky et al. 1994; Sullivan et al. 1994).

Although examining the kinematics of the lower trunk can yield information about how humans coordinate trunk and pelvis movements during gait, the kinematics alone cannot provide a mechanistic explanation for the observed behaviors. Information on muscle involvement in controlling the observed movements is thus required. Past studies have shown that mechanical energy analysis can provide useful information about muscle coordination and motor control (Aleshinsky 1986a–e; Bianchi et al. 1998; Prilutsky and Zatsiorsky 1994; Prilutsky et al. 1996; Winter 1990; Winter et al. 1990). Therefore we also examined the energy transfers across the lower trunk via inverse dynamic analysis of low-back joint moments during gait. We took as our secondary hypothesis that alterations in sagittal plane lower trunk coordination with aging is a result of alterations in flexor/extensor muscle coordination, as reflected by changes in the mechanical energy transfer patterns. The kinematic and kinetic results were then used to discuss behavioral and motor control aspects of lower trunk coordination with aging.

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METHODS

Subjects and gait analysis procedures

The sample consisted of 93 healthy adults (33 males and 60 females) ranging in age from 20.2 to 88.8 yr old (49.1 ± 21.9 yr old, mean ± SD). All subjects were considered healthy at the time of data collection, having no orthopaedic or neurologic disorders affecting locomotion or balance as determined by a staff physician and physical therapist. All subjects provided informed consent prior to inclusion in the study according to institutional policy on human research.

Subjects performed several (range, 1–5; mode, 3) level walking trials at their self-selected speed on a 10-m carpeted walkway in their bare feet. The 10-m walkway provided ample distance for subjects to reach their steady-state gait pattern, normally attained before three steps (~2 m) from a stand still (Miller and Verstraete 1996). Four Selspot II (Selective Electronics, Partille, Sweden) optoelectric cameras were used to track body segment fixed arrays of infrared light-emitting diodes (irLEDs), and two Kistler (Kistler Instruments, Winterthur, Switzerland) piezoelectric force plates were used to record ground reaction forces. Each body segment array consisted of three to five irLEDs imbedded in a flat plastic disk and were securely strapped to the mid-sections of 11 body segments (both feet, shanks, thighs and arms, and pelvis, trunk, and head) using neoprene and Velcro straps. The camera configuration gave an approximate viewing volume of $2 \times 2 \times 2$ m^3 in the center of the 10-m walkway. Force plates were unobtrusively (carpet covered) situated in the center of the viewing volume.

Data were sampled at 150 Hz, and raw kinematic data were filtered using a low-pass Butterworth filter (4th order, 6-Hz cutoff). irLED data were processed using TRACK (Massachusetts Institute of Technology, Cambridge, MA) software to give three-dimensional rotations and translations of each body segment as previously described (Riley et al. 1990). Body segment mass, center of mass, and mass moment of inertia properties were computed from regression equations (McConville et al. 1980) using subject-specific anatomical measurements. For this analysis, only upper body segment data (pelvis, trunk, arms, and head) were required to test our hypotheses. We defined the *trunk* as a rigid segment between the neck joint (C_1–C_2) and *low-back* joint (L_4–L_5), and the *pelvis* as a rigid segment between the low-back joint and the hips. We use the term *lower trunk* to refer to the *trunk–low-back joint–pelvis* as a system.

Upper body segments were defined using a static standing pointing procedure developed by Riley et al. (1990). Briefly, subjects donned the irLED arrays and stood with feet 30 cm apart. During these "static" pointing trials, the tester used two hand-held arrays to point to landmarks that were subsequently used to define each segment's anthropometrics and embedded coordinate system. The low-back joint was defined at the L_4–L_5 level as the mid-point of a line joining the left and right superior iliac crests, and the neck joint was defined at the C_1–C_2 level as the center of a circle circumscribing the neck below the tragus of the ear (Hutchinson et al. 1994; Riley et al. 1990). The segment coordinate system for each body segment, defined from the static pointing trials, were then transformed into the segment's irLED array coordinate system; measurement of the irLED arrays' position and orientation during arbitrary movements enabled estimation of the segments' skeletal kinematics. Precision of array measurements is assessed at 1 mm in translation and <0.1° in orientation (Antonsson and Mann 1989).

Data analysis and key variables

INDEPENDENT VARIABLES: AGE AND GENDER. Two groups of subjects were created from the sample according to age, thus creating two independent variables (age and gender), each with two levels. The majority of young subjects were in their 20s and 30s, and the majority of elderly subjects were in their 60s and 70s: thus a cut-point of 50 yr in age was selected ($n = 49 < 50$ yr old, and $n = 44 \geq 50$ yr old) to ensure the mean age of the groups was adequately separated.

DEPENDENT VARIABLES: PHASE SHIFT AND MECHANICAL ENERGY TRANSFER. Consistent with our hypotheses, primary dependent variables for quantifying lower trunk coordination consisted of *1*) the angular velocity phase angle between the pelvis and trunk (lower trunk phase shift) and *2*) mechanical energy transfer between the pelvis and trunk.

Lower trunk phase shift was determined from peak-to-peak analysis of pelvis and trunk angular velocities within the subject's gait cycle. Because there was only one set of force plates, successive heel strike times for the same foot could not be registered; therefore the gait cycle was determined from peak knee flexion to peak knee flexion of the same limb, depending on which leg (left or right) was visible for both events (Fig. 1). Phase shift was expressed in degrees relative to the gait cycle (1 cycle = 2π radians = 360°). There were typically three peaks in pelvis and trunk angular velocity (Fig. 1): in the early gait cycle (φ_1, negative peaks approximately around heel strike), mid gait cycle (φ_2, positive peaks approximately around mid-late stance), and late gait cycle (φ_3, negative peaks approximately around heel strike of the opposite foot). These three values were averaged to arrive at a mean phase shift (φ) during the cycle. Phase shift angles were computed such that a positive value indicated that the trunk was leading the pelvis.

A top-down inverse dynamic approach was used to compute net

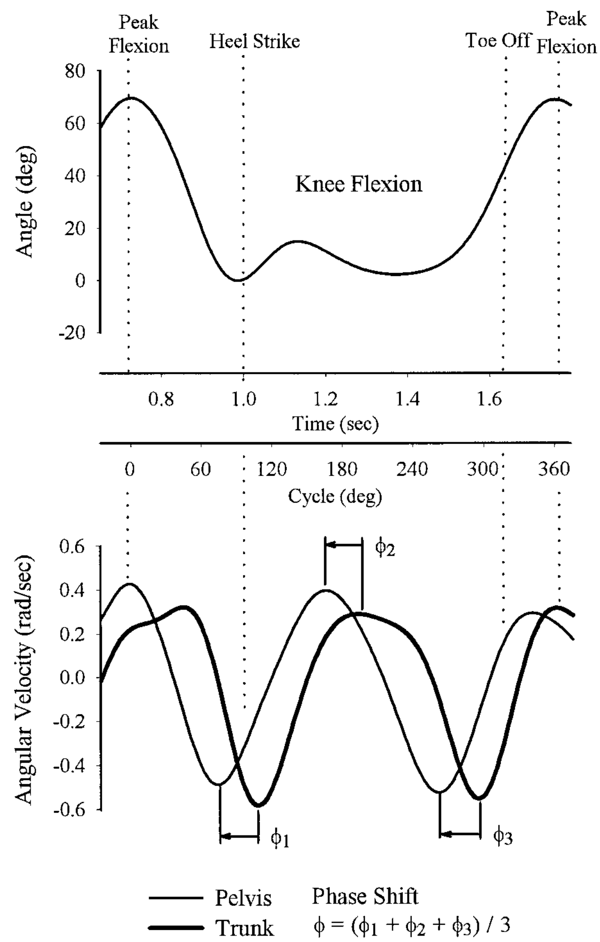

FIG. 1. Gait cycle determination and lower trunk (pelvis-trunk) phase shift quantification. *Top*: knee flexion angle indicating the cycle range (peak knee flexion to peak knee flexion). *Bottom*: pelvis (light line) and trunk (dark line) pitch (sagittal plane) angular velocities during the gait cycle illustrating the peak-to-peak phase angle calculation. Data are from a young female (age, 36.2 yr): note the pelvis leads the trunk by ~30° (1/12 of the gait cycle).

low-back muscle moment during gait trials. Net mechanical power of the low-back in the sagittal plane was then estimated from the expression

$$P_b = M_b^t \cdot (\omega^t - \omega^p) = M_b^t \cdot \omega^t + M_b^p \cdot \omega^p = P_b^t + P_b^p \tag{1}$$

where P_b was the net mechanical power at the low-back joint, P_b^t and P_b^p were the segmental powers (from net muscle moment sources M_b^t and M_b^p, where $M_b^t = -M_b^p$) on the trunk side (proximal) and pelvis side (distal), respectively, of the joint, and ω^t and ω^p were angular velocities of the trunk and pelvis, respectively. The amount of mechanical energy expended was denoted by U_b, and was calculated by integrating the net joint power curve over specific intervals of time (t_1 to t_2, described in more detail below).

$$U_b = \int_{t_1}^{t_2} |P_b| \mathrm{d}t \tag{2}$$

The terms in *Eq. 1* enable important information to be extracted regarding muscle coordination and motor control. *Equation 1* demonstrates that net joint power is a function of the net moment at the joint, and the relative angular velocity of the joint. Therefore when the net moment and angular velocity vectors have the same sense (i.e., both are clockwise or counterclockwise), the net joint power is positive; when the vectors are in opposite directions, the net joint power is negative. As shown by Winter (1990) and others (Eng and Winter 1995; Judge et al. 1996; Prilutsky et al. 1996), a positive net power ($P_b > 0$) means that joint muscles are doing a net amount of positive work; therefore concentric muscle contractions control the observed movements. Conversely, a negative net power ($P_b < 0$) means the joint muscles are doing a net amount of negative work; therefore eccentric muscle contractions control the observed movements. Thus our analysis can separate movements into concentric and eccentric control modes. Because the pelvis-trunk phase shift is a function of trunk and pelvis angular velocities, the direction of phase shift (a leading trunk or a leading pelvis) will affect the sign of the relative angular velocity term ($\omega^t - \omega^p$) in *Eq. 1*, and hence affect the control mode.

However, the energy possessed by a segment is only partially explained by the muscle energy that controls it; energy can also be transferred between segments without the use of muscles (for example, the shank dynamics of an above knee prosthesis during swing phase of gait). The transfer of energy (or flow of power) is determined from the proximal and distal segment powers in *Eq. 1*. As described in detail elsewhere (Aleshinsky 1986a–e), only when both segments rotate in the same direction can there be a transfer of mechanical energy between segments, thereby reducing the burden placed on the muscles for controlling movements. However, no inter-segmental transfer occurs if segments rotate in opposite directions (the muscles do all the work). Thus our analysis also separates movements into transfer and nontransfer modes. Partitioning joint energy expenditures into concentric/eccentric and transfer/nontransfer modes (defined by the time intervals t_1 and t_2 in *Eq. 2*) can therefore be used to quantify the effects of alterations in lower trunk motor control strategies during gait. We chose to analyze three energy expenditure modes: U_b^+, U_b^-, and U_b^o, where superscripts "+" is the energy transfer with concentric action, "–" is the energy transfer with eccentric action, and "o" is the nontransfer with concentric or eccentric action. Figure 2 illustrates mechanical constraints for the two energy transfer modes U_b^+ and U_b^-.

COVARIATES: UPPER BODY POSTURE, RANGE OF MOTION, AND GAIT SPEED. It has been documented in prior reports that upper body posture and range of motion (Sullivan et al. 1994) and walking speed (Himann et al. 1988; Larish et al. 1988) change with age and that walking speed in particular may influence upper body posture (Thorstensson et al. 1984) and the relative phase of the trunk and pelvis (van Emmerik and Wagenaar 1996). Therefore the following variables were documented for use as covariates when testing the main effects of age and gender on lower trunk coordination variables: *1*) peak trunk angle, *2*) peak pelvis angle, *3*) peak low-back angle, *4*) trunk range of motion (ROM), *5*) pelvis ROM, *6*) low-back ROM, and *7*) gait speed. Peak and range were documented for the trunk and pelvis pitch angles (in room coordinates), as well as for the low-back flexion angle (trunk relative to pelvis) during the gait cycle. Gait speed was calculated as the anterior-posterior center of gravity displacement during the gait cycle divided by cycle time and normalized to height.

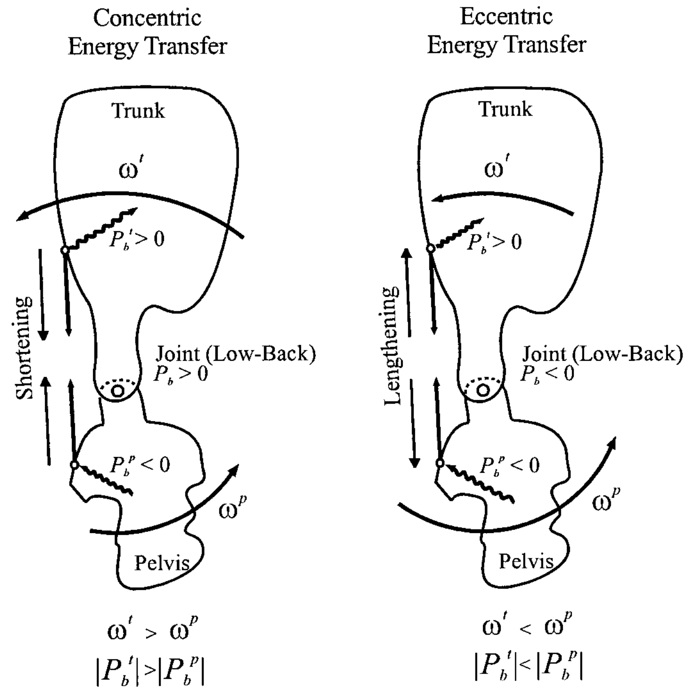

FIG. 2. Schematic diagram of proximal mechanical energy transfer. *Left*: the trunk rotates faster than the pelvis with a net extensor moment; less power flows from the pelvis compared with power flowing into the trunk, thus the muscle must contract concentrically (shortens). *Right*: the trunk rotates slower than the pelvis with a net extensor moment; more power flows from the pelvis compared with power flowing into the trunk, thus the muscle must contract eccentrically (lengthens). Reversal of both pelvis *and* trunk angular velocities will result in a distal flow of power, and reversal of either pelvis *or* trunk angular velocity will result in a no transfer condition.

STATISTICAL ANALYSIS AND HYPOTHESIS TESTING. All variables were evaluated within a single gait cycle and averaged over repeated trials for each subject (73 of the 93 subjects had data for 2 or more trials: 31 had data for 2 trials, 34 had data for 3 trials, 4 had data for 4 trials, and 4 had data for 5 trials). Trial-to-trial repeatability was assessed for dependent variables and covariates using a two-way mixed effects interclass correlation coefficient (ICC, where ICC > 0.80 was considered high repeatability). Subjects who had only one trial were included in the hypothesis testing, but excluded from repeatability tests. Associations between variables were evaluated with Pearson's product moment correlations, and between-groups differences were evaluated with two-way ANOVA using age and gender as independent variables. Covariates influences were assessed with partial correlation analysis and two-way analysis of covariance (ANCOVA). Interaction effects and homogeneity of variance were assessed for each main effects test. An alpha level of 0.05 was chosen for all statistical tests. SPSS for Windows (Version 8.0, SPSS, Chicago, IL) was used for all statistical analyses.

Procedures for testing hypotheses were as follows. First, main effects analysis was performed for covariates (treating them as dependent variables), as well as correlation analysis among covariates and between the covariates and age. Hypothesis one was then tested by two-way ANOVA (phase shift as dependent variable, and age and gender groups as the independent variables), and then by ANCOVA (gait speed or upper body angles as covariates) to assess whether the covariates explained the ANOVA main effects differences. Pearson correlations and partial correlations were also performed (with age as a continuous variable) as a secondary approach for testing hypotheses. Hypothesis two was tested similar to hypothesis one (except mechanical energy expenditures were the dependent variables).

RESULTS

Subject characteristics and covariates

Young subjects were not significantly different in height ($P = 0.095$) and weight ($P = 0.948$) compared with old subjects (Table 1). There were nonsignificant age differences ($P = 0.098$) between males and females; however, males were taller ($P < 0.001$) and heavier ($P < 0.001$) than females (Table 1).

Young subjects walked faster per unit height ($P = 0.007$) compared with old subjects, and females walked faster per unit height ($P = 0.007$) compared with males (Table 2). Low-back flexion ROM was significantly greater in young subjects compared with old subjects ($P = 0.021$) but not significantly different ($P = 0.893$) between males and females (Table 2). There were no significant differences between old and young subjects or between males and females in peak trunk pitch angle ($P = 0.470$ and $P = 0.230$), peak pelvis pitch angle ($P = 0.370$ and $P = 0.639$), peak low-back flexion angle ($P = 0.953$ and $P = 0.912$), trunk ROM ($P = 0.096$ and $P = 0.242$), and pelvis ROM ($P = 0.453$ and $P = 0.190$).

Correlation analysis indicated weak-to-moderate significant relationships between gait speed and age ($r = -0.366$, $P < 0.001$), low-back ROM and age ($r = -0.256$, $P = 0.013$), and between low-back ROM and gait speed ($r = 0.267$, $P = 0.010$). When controlling for gait speed, ANOVA indicated no significant difference between young and old subjects in low-back ROM ($P = 0.095$). Because gait speed and low-back ROM were statistically different between gender and/or age groups, and had statistically significant associations with age, they were used as covariates in the main effects analysis of the primary dependent variables.

Trial-to-trial repeatability for these variables was found to be high (ICC = 0.970 for gait speed and ICC = 0.853 for low-back flexion ROM). The homogeneity of variance test was not violated (Levene's test, $P > 0.05$) for any of the above main effects tests, and there were no significant interactions ($P > 0.05$) in the two-way ANOVA tests.

Lower trunk phase shift

Lower trunk phase shift for age and gender groups are summarized in Table 3. Average phase shift (φ) was significantly different between young and old subjects ($P < 0.001$) but not different between males and females ($P = 0.452$). On average, young subjects lead trunk angular motion with their pelvis, while old subjects lead pelvis angular motion with their trunk. Phase shift was significantly associated with age ($r = 0.480$, $P < 0.001$), gait speed ($r = -0.287$, $P = 0.005$) and low-back ROM ($r = -0.227$, $P = 0.029$). Figure 3 shows the relationship between phase shift and age.

To determine whether this relationship was controlled by gait speed and/or low-back ROM, partial correlation and analysis of covariance tests were conducted between phase shift and age controlling for gait speed and low-back ROM. Gait speed accounted for only a small proportion of the association between lower trunk phase shift and age; when controlling for gait speed the correlation remained significant (phase vs. age partial $r = 0.420$, $P < 0.001$), and the difference between young and old subjects remained significant ($P < 0.001$). Furthermore, controlling for low-back ROM was even less explanatory (phase vs. age partial $r = 0.448$, $P < 0.001$), which is not surprising due to the apparent dependence of low-back ROM on gait speed.

TABLE 1. *Subject characteristics*

	Age Group*		Gender	
	Young	Elderly	Male	Female
Age, yr	29.80 ± 6.81	70.67 ± 8.66	51.45 ± 21.10	47.87 ± 22.42
Height, m	1.68 ± 0.09	1.65 ± 0.10	1.74 ± 0.07	1.62 ± 0.08
Weight, kg	67.71 ± 13.96	68.51 ± 15.54	79.47 ± 13.68	61.83 ± 11.01

Values are means ± SD. Number of Young subjects is 49, Elderly is 44, Male is 33, and Female is 60. * Young subjects, <50 yr old; Elderly subjects, >50 yr old.

TABLE 2. *Covariates: gait speed and upper body angles (pelvis, trunk, and trunk relative to pelvis) kinematics during preferred speed gait*

	Age Group[a]		Gender	
	Young	Elderly	Male	Female
Gait speed[b]	1.26 ± 0.15	1.13 ± 0.20	1.18 ± 0.16	1.20 ± 0.20
Pelvis peak[c]	5.54 ± 2.31	6.06 ± 3.34	5.98 ± 2.99	5.68 ± 2.77
Trunk peak	5.48 ± 2.06	6.12 ± 4.04	5.27 ± 2.47	6.07 ± 3.46
Low-back[d] peak	7.32 ± 3.50	7.43 ± 4.81	7.31 ± 4.50	7.40 ± 3.98
Pelvis ROM[e]	4.20 ± 1.38	4.38 ± 1.52	4.55 ± 1.71	4.14 ± 1.27
Trunk ROM	4.37 ± 1.18	3.99 ± 1.61	3.97 ± 1.29	4.31 ± 1.46
Low-back ROM	4.78 ± 1.80	3.95 ± 1.68	4.35 ± 1.94	4.41 ± 1.70

Values are means ± SD. Number of Young subjects is 49, Elderly is 44, Male is 33, and Female is 60. [a] Young subjects, <50 yr old; Elderly subjects, >50 yr old. [b] Gait speed in m/s (not corrected for height in Table). [c] Peak pitch angle (deg) during gait cycle. [d] Low-back = trunk relative to pelvis at L_4–L_5 joint. [e] Range of motion (ROM) of pitch angle (deg) during gait cycle.

TABLE 3. *Dependent variables: lower trunk phase shift angle, and lower trunk joint mechanical energy expenditures during preferred speed gait*

	Age Group[a]		Gender	
	Young	Elderly	Male	Female
Phase angle,[b] φ	−10.86 ± 20.21	6.66 ± 19.42	−0.17 ± 20.93	−3.89 ± 22.03
Concentric,[c] U_b^+	0.48 ± 0.29	0.36 ± 0.50	0.49 ± 0.35	0.39 ± 0.43
Eccentric,[d] U_b^-	0.35 ± 0.35	0.62 ± 0.77	0.58 ± 0.84	0.42 ± 0.41
No transfer,[e] U_b^o	1.28 ± 1.68	0.83 ± 1.25	1.14 ± 1.37	1.03 ± 1.58

Values are means ± SD. Number of Young subjects is 49, Elderly is 44, Male is 33, and Female is 60. [a] Young subjects, <50 yr old; Elderly subjects, >50 yr old. [b] Lower trunk phase shift (degrees): positive = trunk leads pelvis, negative = pelvis leads trunk. [c] Energy expenditure of low-back joint during concentric energy transfer (100 ∗ Joules/kg). [d] Energy expenditure of low-back joint during eccentric energy transfer (100 ∗ Joules/kg). [e] Energy expenditure of low-back joint during no energy transfer (100 ∗ Joules/kg).

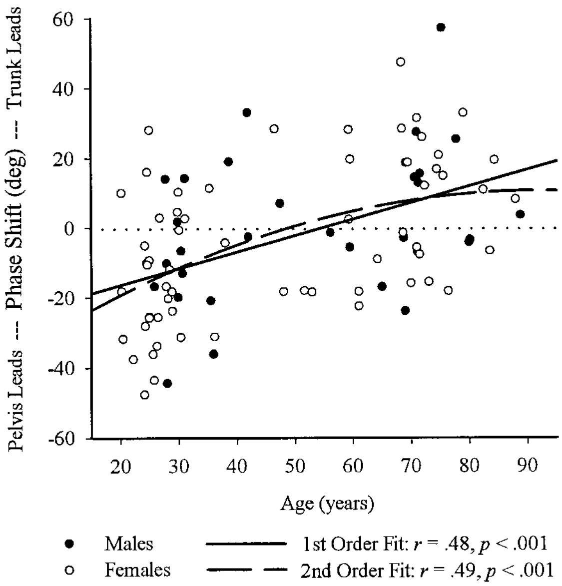

FIG. 3. Scatter plot of lower trunk phase shift vs. age. The solid line is a linear regression fit (1st order) through all points. Females are indicated by open circles and males by closed circles. The dashed line is a quadratic regression fit (2nd order) through all points, indicating a threshold (no further increase) for the phase shift after approximately 80 yr old. The 1st-order regression line intersects the zero phase shift line at approximately 55 yr old, while the 2nd-order regression line intersects at approximately 49 yr old.

Repeatability was assessed for the three (φ_1, φ_2, and φ_3) peak-to-peak phase delay measures and was found to be high (ICC = 0.854). Trial-to-trial repeatability for average pelvis-trunk phase shift angle was also found to be high (ICC = 0.911). The homogeneity of variance test was not violated (Levene's test, $P > 0.05$) for the main effects test, and there were no significant interactions ($P > 0.05$) in the two-way ANOVA or ANCOVA tests.

Low-back mechanical energy transfer

Mechanical energy expenditures of the low-back are summarized in Table 3. Analysis of variance indicated that older subjects transferred more eccentric muscle energy, U_b^-, compared with younger subjects ($P = 0.013$). Younger subjects transferred more concentric muscle energy, U_b^+, and expended more energy without transfer, U_b^o, but these differences were not significant ($P > 0.05$). There were no significant differences by gender ($P > 0.05$). A similar result was obtained when controlling for gait speed. However, when controlling for lower trunk phase shift, the differences between young and old subjects for U_b^- disappeared ($P = 0.157$), while the differences between young and old subjects for U_b^+ and U_b^o registered significant ($P = 0.008$ and $P = 0.028$, respectively). The effect of the lower trunk phase shift on low-back power flow is demonstrated in Fig. 4 for a representative young (age = 36.2 yr) and elderly (age = 74.8 yr) subject: the phase shift reversal causes a reversal in the sense of the power curve.

Trial-to-trial repeatability of the mechanical energy data were found to be high (ICC = 0.915 for U_b^+, ICC = 0.896 for U_b^-, and ICC = 0.922 for U_b^o). There were no significant interactions ($P > 0.05$) in the two-way ANOVA or ANCOVA tests. The homogeneity of variance test was not violated (Levene's test, $P > 0.05$) for any of the above main effects tests, except for U_b^- that did show a higher error variance in elderly subjects compared with younger subjects. The effects of this violation are assumed small due to the similar group sizes for young and elderly.

DISCUSSION

Our data suggest that a reversal in the angular velocity phase relationship of the trunk and pelvis occurs with aging in asymptomatic adults that is independent of walking speed and range of motion of the low back. Furthermore, this reversal in phase relationship parallels an apparent reversal of the muscle lengthening and shortening (eccentric and concentric) sequences of low-back flexor-extensor musculature, as indicated by the mechanical energy transfer patterns of the low back. Thus there appears to be two distinctive lower trunk coordination strategies used by individuals during preferred-speed gait (trunk-leading and pelvis-leading), where elderly subjects appear more likely to use a trunk-leading strategy than young subjects. Our data also suggest that elderly subjects increase the mechanical energy demands of their low-back musculature, and that this increase is a consequence of using the trunk-leading strategy.

Effects of aging on lower trunk coordination during gait

We found that the correlation between age and pelvis-trunk angular velocity phase angle (phase shift) was moderately strong ($r = 0.480$, $P < 0.001$) with the regression line crossing the zero phase line at approximately 55 yr of age. The predominant lower trunk coordination strategy for younger subjects was to lead their trunk with the pelvis (mean of $-10°$), while the predominant strategy for elderly subjects was to lead their pelvis with the trunk (mean of $+7°$). Of particular interest was that the observed age-related differences in lower trunk coordination did not appear to be mediated by forward velocity of the center of mass or upper body posture: lower trunk phase shift correlated with gait speed and low-back ROM, but controlling for those variables did not have a large effect on the relationship between lower trunk phase shift and age.

Although we did detect an age- (but not gender) related difference in low-back ROM, the difference between young and old subjects was small (~1°) and could be statistically explained by differences in walking speed. These data indicate that the upright posture of the trunk was similar for young and old, and males and females, in this study. Furthermore, magnitudes and range of trunk ROM were similar to data reported by others (Krebs et al. 1992; Murray 1967; Opila-Correia 1990; Stokes et al. 1989; Thorstensson et al. 1984). Age and gender differences were found in gait speed, and there was a significant association between age and gait speed. Van Emmerik and Wagenaar (1996) showed that within healthy young individuals, the relative phase between trunk and pelvis increased with faster walking. It should be noted that van Emmerik and Wagenaar (1996) computed continous phase and discrete phase relationships between the pelvis and trunk based on angular displacements in the transverse plane. Nonetheless, in our study, the relationship between gait speed and phase shift was consistent with van Emmerik and Wagenaar (1996), although the variation in gait speed across our subjects did not account for much of the variance in phase shift (~8%), prob-

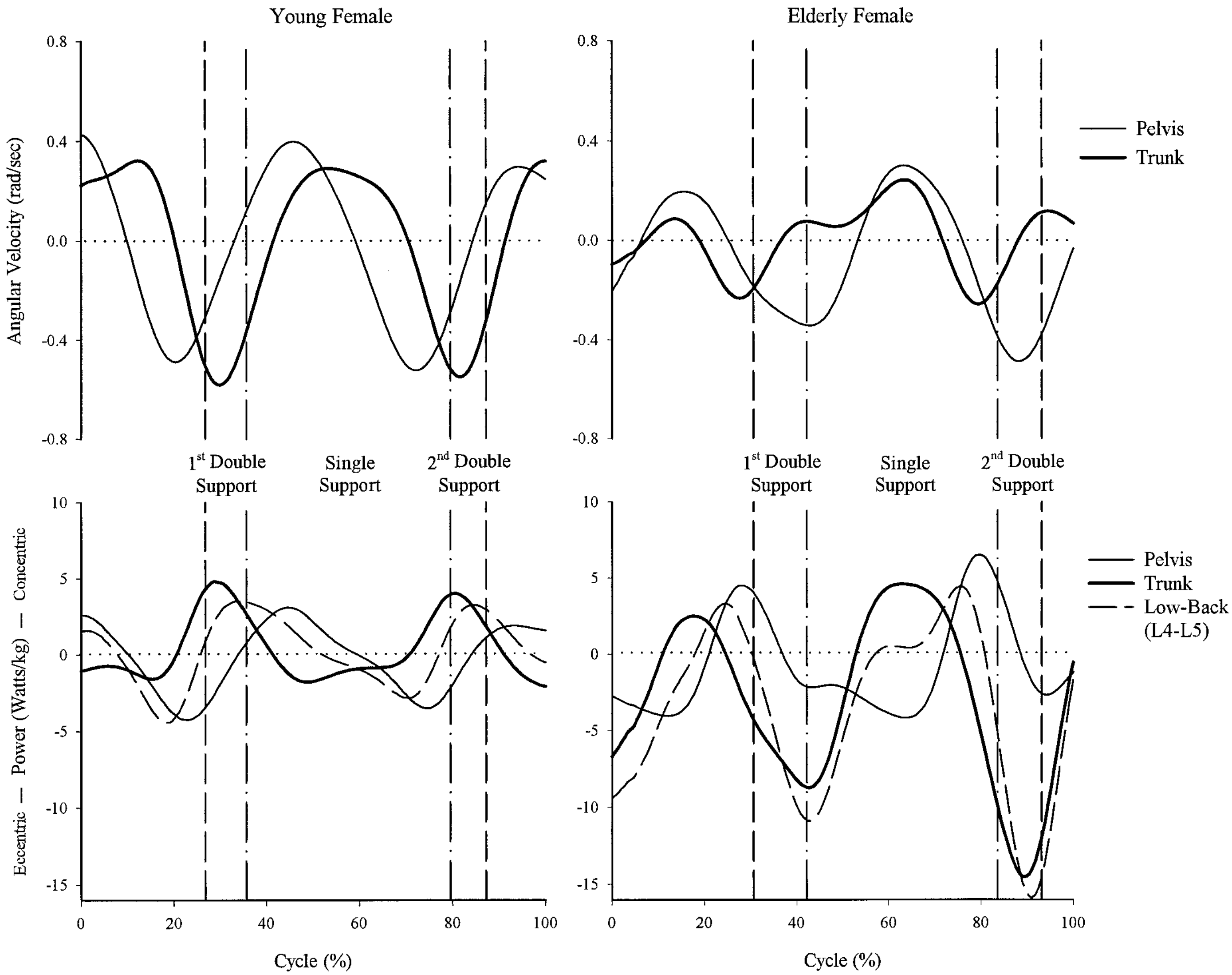

FIG. 4. Representative data for a young female (age, 36.2 yr) subject and elderly female (age, 74.8 yr) subject. *Top*: angular velocities of pelvis (thin solid line) and trunk (thick solid line) indicating pelvis leading trunk for the young subject and trunk leading the pelvis for the elderly subject. *Bottom*: net mechanical power at the low-back (thick solid line) and the segmental power components, proximal pelvis (thin dashed line) and distal trunk (thin solid line), indicating the reversed sense of the net power curve due to the phase delay reversal of the pelvis and trunk.

ably because subjects in our study walked at only their preferred speed. Although walking speed is a commonly used measure of gait performance in elderly subjects (Himann et al. 1988; Larish et al. 1988; Oberg et al. 1993; Ostrosky et al. 1994), our results suggest that gait speed is not appreciably affected by different strategies of lower trunk coordination. This may be important if one considers the potential consequences of the lower trunk coordination strategy used.

A gait style in which the pelvis leads the trunk would suggest a predominant lower extremity controller schema; the CNS is using the lower body to direct movements of the upper body. Conversely, a gait style in which the trunk leads the pelvis would suggest a predominant upper body controller schema; the CNS is using the massive upper body to direct movements of the lower body. It stands to reason that the trunk-leading coordination strategy may impact the ability to recover from a trip or slip. Adequate lagging of the trunk behind the pelvis would allow more time for the motor cortex to process sensory information and issue corrective commands to stabilize the upper body when the path of the lower extremities is disturbed. Leading with the massive trunk may not allow a sufficient upper body stabilizing recovery response. Because gait speed explained only a small portion of the age-phase shift relationship, gait speed may not always be the best predictor of falls risk. Pavol et al. (1999) showed that elders who walk faster are less likely to recover from a trip and that trunk orientation was not a good predictor of falls recovery. They (Pavol et al. 1999) suggested that available power reserves do not affect the ability to recover from fall. Our data offer an explanation; the ability to generate rapid torque (power burst) may not be sufficient for trip recovery response when the trunk leads the pelvis by a substantial margin. Future studies should attempt to quantify the influence of lower trunk coordination on falls recovery, and determine specifically what (if any) magnitude of trunk lead constitutes a threat to successful trip recovery.

The low-back joint power profiles in Fig. 4 suggest that elders contract low-back muscles (flexor-extensor) eccentrically in the double support phases and early single support phase and concentrically in late single support phase, while younger subjects do the opposite. The trunk-leading strategy used by elders resulted in an increase in the mechanical energy expenditure of the low-back musculature; elderly subjects expended more energy in eccentric control than younger subjects, a difference that was explained by differences in lower trunk phase shift. This finding suggests that elderly subjects rely more on eccentric control of low-back muscles to regulate energy transfer during the double limb support phase of gait (Fig. 4), which implies an alteration in activation strategies of the nervous system (Enoka 1996). When accounting for the phase shift differences among subjects, the younger subjects used more concentric muscle power compared with elderly subjects, suggesting that elders reduce the concentric energy output of trunk muscles to minimize the energy transferred proximally to the trunk during the less dynamically stable single support phase of gait.

Behavioral and motor control aspects of lower trunk coordination strategies

These data raise some important questions: Why would elders select an upper body, trunk-leading, control strategy, particularly given its potentially negative consequences? What is the mechanism that enables the trunk-leading strategy to occur without appreciably affecting upper body posture or walking speed in healthy adults?

Eccentric muscle activity is not as metabolically demanding as concentric muscle activity (Willems et al. 1995), and hence may be a compensatory pattern adopted by elders to minimize low-back muscle fatigue. However, the mechanisms of muscle fatigue under low contraction force conditions are not well understood, and recent evidence suggests that neural adaptations following immobilization of muscles (which might translate into increased sedentary lifestyle of the elderly) may actually increase muscle endurance (Semmler et al. 2000), particularly in women (Semmler et al. 1999). It is known, however, that prolonged eccentric activity generates a higher concentration of plasma creatine kinase that can induce significantly prolonged muscle pain compared with that caused by concentric activity (Lund et al. 1998), and therefore eccentric efficiency is not a satisfactory explanation. It is also likely that strength deficits of the lower extremities are, at least, a contributing factor. Previous gait analysis studies have shown increased hip power in healthy elders compared with young subjects (Judge et al. 1996), and increased hip and low-back power in disabled elders compared with healthy elders (McGibbon et al. 2001), suggesting that a trunk-leading strategy may be a response to lower extremity impairment. In essence, these past studies suggest that pelvis and trunk musculature is used to advance, or pull, the leg into swing phase when the lower extremity muscles (ankle plantar flexors and knee extensors) are weakened from aging and/or disability. Although strength was not measured in our sample, past studies support our expectation that our elderly subjects were weaker than our younger subjects (Frontera et al. 1991).

What facilitates this compensatory response? The firing sequences for leg muscles in automated movements such as gait are, at least in animals (Golubitsky et al. 1999), issued via central pattern generators (CPGs). It has been proposed that the functional organization of CPGs constitutes a network of coupled oscillators that produce coordinated multi-joint movements (Grillner et al. 1995; Pearson 1993). Experimental data have shown that the phase coupling of kinematic patterns of the lower extremity segments are highly invariant at different walking speeds (Bianchi et al. 1998) and with different trunk postures (Grasso et al. 2000). Walking backward has been shown to preserve the kinematic profiles of segmental angle elevations but with a reversal in the phase coupling of limb segments and patterns of muscle activity (Grasso et al. 1998). Grasso et al. (1998) and Grillner (1981) hypothesized a reorganization of motor programming such that the CPGs recruited for backward movements were the same series of coupled oscillators but with a sign reversal of the phase coupling.

Although theories surrounding the role of CPGs and their interaction with high- and mid-level control centers of the brain remain controversial (Mitz and Winstein 1993), the reversal in lower trunk phase without a statistically significant alteration in gait speed and upper body posture (and in the presence of normal vestibulocerebellar function) suggests that reorganization of lumbo-sacral muscle phasic oscillators may occur in response to lower extremity weakness (or other impairments) in otherwise asymptomatic elderly adults, to maintain proper attitude control of the trunk and overall gait function. More detailed studies that use electromyography would be required to test the above hypothesis.

Limitations and conclusions

Studies of locomotion in aging individuals are confounded by aging effects of all body systems including neurologic, musculo-skeletal, cardiac, and respiratory systems. Although the subjects in our study were healthy and free of orthopaedic, cardiac-respiratory, and neurologic disorders, we cannot preclude that subtle age-related changes in multiple body systems influenced their locomotor abilities. We only examined the mechanics of the sagittal plane, and therefore it is unknown what age-related changes, if any, occur in the frontal and transverse planes. Our model of the upper body consisted of rigid segment models of the pelvis, trunk, arms, and head; treatment of the trunk and perhaps the arms as multi-jointed segments may also yield interesting information. It should also be understood that mechanical energy analysis, based on inverse dynamic models, provides only indirect muscle action data. Energy transfer calculations for the low back are a function of low-back net moments, which can be calculated top-down or bottom-up. Although there are no published studies that suggest one method is better, preliminary comparisons from our lab show very close agreement between the two methods (unpublished observations); a top-down approach was chosen here to enable calculation of lower trunk energy transfers without the use of force plates.

In conclusion, our data suggest that lower trunk coordination strategy during gait may be altered in humans of advanced age and is likely a response to subtle age-related changes in locomotor function. Future studies comparing lower trunk coordination between healthy subjects and patients with neurologic (e.g., cerebellar disease, Parkinsons, and spinal cord injury) and musculoskeletal (e.g., arthritis, low-back pain, and muscle wasting) dysfunction are warranted and may shed light on this previously unreported characteristic of gait with aging. The behavioral and motor control aspects of these findings may be important for understanding locomotor impairment in aging humans and in quantifying falls risk.

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