VA RESEARCH & DEVELOPMENT

JRRD

Journal of Rehabilitation Research & Development

Vol. 36 No. 3, July 1999

Effects of age and functional limitation on leg joint power and work during stance phase of gait

Chris A. McGibbon, PhD and David E. Krebs, PhD, PT

Massachusetts General Hospital, Department of Orthopaedics, Biomotion Laboratory, Boston, MA 02114; MGH Institute of Health Professions, Boston MA 02114

Abstract--It is commonly accepted that leg muscle power is an important component of functional ability. Paced gait data for 20 healthy young women (27±4.2 yrs), 16 healthy old women (72.5±5.6 yrs), and 24 functionally limited old women (73.5±7.2 yrs) were analyzed during stance phase to investigate whether power calculations from motion analysis data were sensitive to impairments beyond the differences expected with aging. Healthy women included in the study functioned at a high/moderate level with no limitations, while functionally limited women functioned at a much lower level and exhibited one or more functional limitations as defined by the SF36 physical function scale. Single support time (SSUP), stance duration (SDUR), average forward center of gravity velocity (GVEL), and three-dimensional net power and work of the ankle (APOW/AWRK), knee (KPOW/KWRK), and hip (HPOW/HWRK) were computed for the stance limb during the stance phase of gait. Univariate ANCOVA was used to examine which variables were most sensitive to functional limitations. We found that SDUR and SSUP were not different among the three groups when controlling for height and weight. Although differences in CG velocity between healthy and functionally limited old women were not significant, both elderly groups translated their CG slower than did the younger women ($p<0.0001$) when walking at the same cadence. Controlling for CG velocity eliminated all significant differences among groups except for APOW and AWRK in late stance phase between healthy and functionally limited elderly women ($p<0.003$). These results suggest that decreased ankle plantar-flexor power in late stance of gait may be an impairment-related characteristic more than an age-related characteristic.

Key words: *body size, elderly, gait, mechanical power and energy, velocity.*

INTRODUCTION

Distinguishing between mobility characteristics caused by impairments and those typical of aging is a critical, but rate-limiting step in determining the effectiveness of interventions intended to prevent disablement. Physical, social, and psychological factors of aging, such as strength and balance deficiencies (1-3), malnutrition (4,5), medications, and depression (6) can limit walking ability and increase the risk of falling (7-11). Time-distance and kinematic characteristics of elderly gait have been extensively reported (12-18). Decreases in gait velocity and relative increases in stance duration and double support time with increasing age probably reflect a trend toward a safer, more dynamically stable, gait pattern (18,19) and are clearly consequences of the aging process, but not necessarily indicative of functional limitations. Brown et al. (20) have suggested that muscle and link-segment power analyses may be more sensitive to functional limitations than time-distance or kinematic measures. There is evidence that loss of muscle power decreases more rapidly with age than loss of muscle strength (21). The ability to generate rapid movements to recover from dynamic instability may be dependant upon the absolute power generating capacity of the muscles involved (22). Several investigators (23-25) have reported a strong relationship between isokinetic leg power and functional performance measures (timed chair rise, stair climbing, lifting, and walking). Whether gait analysis-based net muscle power and work calculations are sensitive to functional limitations, however, has not been thoroughly examined.

Few studies have examined the power and work characteristics of the lower limbs during functional activities, such as gait, in an elderly population (19,26). Comparisons made between healthy young and fit elders during gait have not revealed remarkable differences (19,26) that cannot be explained by common elderly gait characteristics, such as decreased gait velocity or step length (26). Mechanical power, the rate of change of energy due to mechanical sources, is the scalar product of force and velocity ($F \cdot v$), or torque and angular velocity ($T \cdot \omega$). Strictly speaking, muscle power implies a muscle force multiplied by its shortening (or lengthening) velocity. However, inverse dynamic analysis of human movement provides only the net torque and net force acting at the joints, and not muscle forces. The linear power component ($F \cdot v$) cancels at the joint; therefore, the net power across the joint is angular ($T \cdot \omega$) only. Therefore, we define net muscle power as the instantaneous angular power component of muscle torque, and net muscle work as the time integrated value of net muscle power.

From an analytical point of view, calculations of net muscle power and work are highly dependent upon body size (mass, stature) and the kinematics (velocity) of movement. Because elders, in general, walk more slowly than do young adults, a reduction in muscle power generated (and absorbed) would be expected. Differences in height and weight between young and elderly subjects (27) may also influence muscle power magnitudes. It is also unclear whether net muscle power or the net muscle work is better for distinguishing between unimpaired and impaired walkers. Certainly, arguments can be made for both sides, although there is no empirical evidence available to warrant choosing one analysis technique over another.

We calculated the net muscle power and work of the leg joints (ankle, knee, and hip), and selected time-distance parameters (center of gravity velocity, stance duration, and single support time), in three groups of women: healthy young women, healthy elderly women, and functionally limited elderly women, during the stance phase of moderate-speed paced walking. The purpose of our investigation was twofold: to determine if net muscle power and work, and selected time-distance parameters, were sensitive to functional limitations when the effects of body size and forward velocity are accounted for; and to compare the use of net muscle power and net muscle work as a means of predicting age and impairment-related differences in gait. Our findings may assist future investigations of walking impairment and the effects of intervention on functional mobility of the elderly.

METHODS

Leg joint mechanical power was analyzed for three groups of women, totaling 60 subjects. Groups consisted of 20 healthy young women (27.0±4.2 yrs), 16 healthy elderly women (72.5 ±5.6 yrs), and 24 functionally limited elderly women (73.5±7.2 yrs). General inclusion criteria for the present study required that all subjects successfully complete at least one 120 beat-per-minute (BPM) paced gait trial during their testing protocol. All subjects gave signed consent in accordance with institutional policy regarding research on human subjects.

Healthy Subjects

Healthy young women and healthy elderly women were included if they were free of any neurologic and orthopaedic disorders. The young women were selected if they were within the age range of 20 and 35 years. The healthy elderly women were further screened by telephone questionnaire regarding participation in mild, moderate, and vigorous daily activities at home, work, and recreational activities. The healthy elderly were selected for the study if they functioned at a high level with no limitations, and were between the ages of 60 and 90 years. Subjects were excluded if they had any cardiac history, neurological diseases, recent cancer with ongoing medical treatment, uncontrolled hypertension, diabetes or seizures, recent fractures, or were legally blind. Subjects were also excluded from the healthy elderly group if they regularly used an assistive device for ambulation.

Functionally Limited Subjects

The functionally limited subjects were selected from a larger population of 133 community dwelling elders enrolled in a separate study to evaluate functional ability and disability status before and after a strengthening intervention (28). The criteria for inclusion were as follows:

- the ability to use the telephone
- the ability to ambulate indoors with or without an assistive device but not requiring human assistance
- being cognitively intact
- having permission from one's primary care physician
- reporting some limitation in at least 1 of the 9 functional areas (excluding vigorous

activity) listed in the SF36 physical function scale[1].

These subjects were included even if they had disorders, such as osteoarthritis, that contributed to their functional limitations. Exclusion criteria for the functionally limited elders were the same as for healthy elders with the exception of ambulation with assistive device. Only baseline (first visit) data for the functionally limited subjects were analyzed in the present study. Inclusion of gait data for the functionally limited subjects also required that their combined isometric leg muscle strength (knee extensor + hip extensor + hip abductor) be less than 26 kg_f; thus, representing the weakest women in a population of 133 functionally limited elders. Disability status for these subjects is reported by Jette et al. (28).

Muscle strength and disability status could not be compared among groups because the healthy subjects were recruited for other research projects and different instruments were used to evaluate these variables. However, given the strict inclusion/exclusion criteria above, the functionally limited subjects functioned at a much lower level than did the healthy elderly subjects.

Gait Analysis and Time-Distance Parameters

Rigid plastic disks consisting of an array (3 or more) of light-emitting diodes (LEDs) were strapped to the midsection of 11 body segments: feet, shanks, thighs, pelvis, trunk, arms, and head. Subjects walked along a 10-m walkway in bare feet while keeping their pace to a metronome set at 120 BPM. Other than cadence, no other constraints were imposed on subjects' gait. Bilateral LED array trajectories were measured with four Selspot II (Selective Electronics, Partille, Sweden) optoelectric cameras, from which six degree of freedom (DOF) body segment kinematics (position and orientation) were calculated. Two piezoelectric force plates (Kistler Instruments, Winterthur, Switzerland) were used to measure foot-floor reaction forces. Camera and force plate data were synchronously sampled at 150 Hz and raw camera data were smoothed with a low pass filter (6 Hz cutoff frequency). Inertial parameters of body segments (mass, center of mass, and mass moment of inertia) were computed on a subject-specific basis using anatomical measurements and regression equations for women (29) and joint angles were calculated from Cardan 3-1-2 decomposition of the segment-to-segment rotation matrices (30).

Whole body center of gravity (CG) velocity (GVEL) was calculated as the average anterior-posterior velocity of the CG between heel strike and toe off. Stance duration (SDUR) was determined from the force plate vertical reaction trace: heel strike was assumed to occur when the vertical reaction force was greater than 20 N (sensitivity of the vertical force channel of the Kistler force plate was 10 N), and toe-off time was assumed to occur at the first frame following heel strike where the vertical force was less than 20 N. Because the contralateral (swing) leg was not, in general, contacting a force plate during ipsilateral stance, and foot switches were not used for these subjects, single support time (SSUP) was assumed to occur between the first and second maxima of the vertical force plate trace (31).

Net Muscle Power and Work

Six DOF segment kinematics, force plate data, and inertial parameters were input into a Newtonian inverse dynamic algorithm (32) to compute the instantaneous three-dimensional (3D) net joint moments at the ankle, knee, and hip of the stance leg from heel strike (zero percent stance) to toe-off (100 percent stance). Higher order kinematics (velocities and accelerations) were calculated numerically using a Lagrangian 5-point differentiation algorithm. The net muscle power (*P*) at each joint (ankle, knee, and hip) was expressed as the product of net joint moments and joint angular velocity in segment coordinates (ankle power in shank coordinates; knee power in thigh coordinates; and hip power in pelvic coordinates). Net muscle power for the *j*th joint was written

[1]

$$P_j = T_j \cdot \omega_j = (T_x\omega_x + T_y\omega_y + T_z\omega_z)_j$$

where *T* denotes the net muscle torque and ω denotes the angular velocity of joint *j*; *z* denotes the axis normal to the segment's sagittal plane, *x* denotes the axis normal to the segment's coronal plane, and *y* denotes the long axis of the segment. The mechanical work (*U*) at each joint was calculated by integrating selected positive (+) and negative (-) intervals of the power curves, which are described below.

[2]

$$U_j^{(+)} = \int_a^b P_j^{(+)} dt \quad , \quad U_j^{(-)} = \int_a^b P_j^{(-)} dt$$

Data Analysis

Three-dimensional analysis of three leg joints generates nine joint power curves. Because the differences in power between subjects may be at certain intervals of the stance phase, and not at others, it was desirable to analyze various positive (generative) and negative (absorptive) portions of each power curve. To limit the number of variables in the analysis, we chose to analyze ankle plantar- dorsiflexion, knee flexion-extension, hip flexion-extension, and hip abduction-adduction. As shown by Eng and Winter (33), energies developed in the sagittal plane for leg joints, and frontal plane of the hip joint, were substantially greater than those developed in other leg joint planes. **Figure 1** shows the portions of each joint power curve analyzed. They are described as follows:

- for the ankle - the absorptive portion of the power curve in mid stance (A1), and the generative portion in late stance (A2)
- for the knee - the absorptive portions in early (K1) and late stance (K3) and the generative portion in mid stance (K2)
- for the sagittal plane hip - the generative portions in early (HS1) and late stance (HS3) and the absorptive portion in mid stance (HS2)
- for the frontal plane hip - the absorptive portion in early stance (HF1) and the generative portions in late stance (HF2).

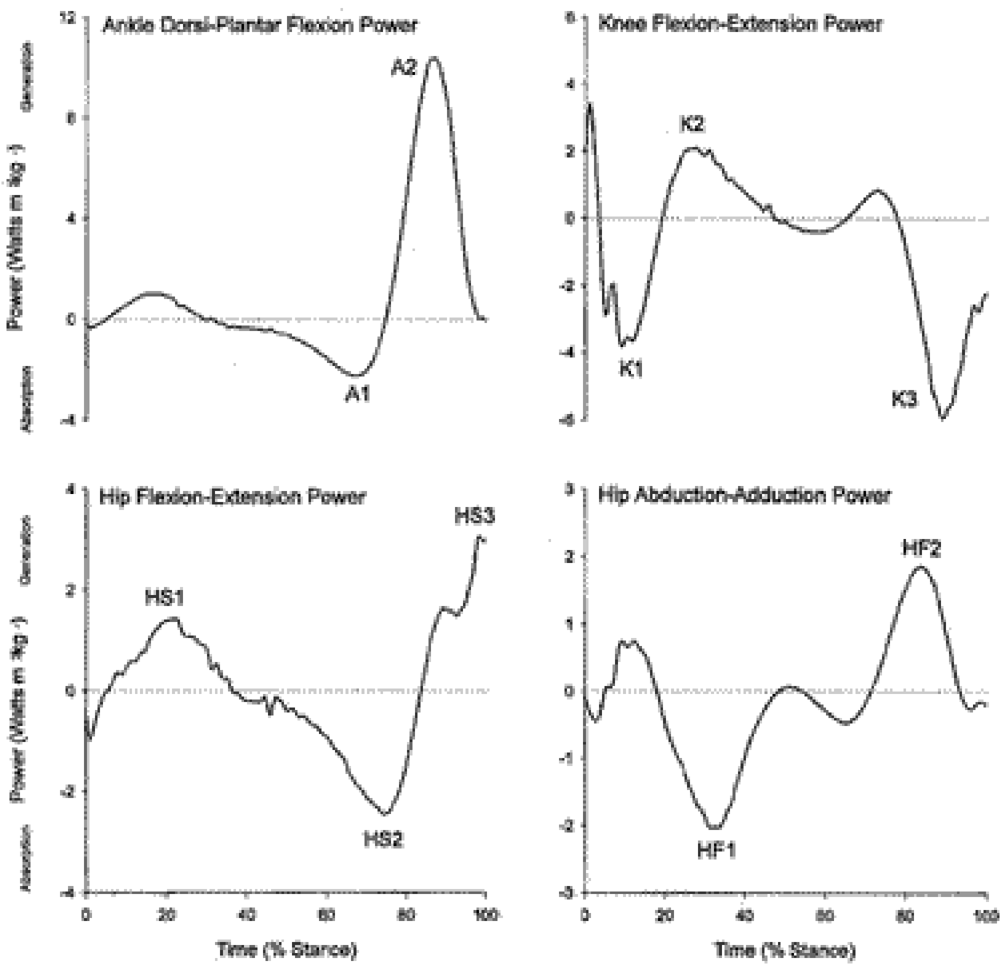

Figure 1.
A representative subject's data used to show the generative (positive) and absorptive (negative) portions of the power curve analyzed in this study.

Because power is instantaneous, the peak power value during any given time interval was used to represent *net muscle power* and the time integrated power during that time interval was used to represent *net muscle work*. Three separate analyses were performed. The first analysis examined between-group differences in peak net muscle power at the above portions and the second analysis examined between-group differences in net muscle work over the duration of each portion. The third analysis compared the time-distance measures between groups.

Mechanical power and work data were normalized to body mass index (Watts m^2 kg^{-1} and Joules m^2 kg^{-1}, respectively) and CG velocity was corrected for body height. Analysis of covariance (ANCOVA) was used to determine between-group differences in power and work using CG velocity as a covariate, and between-group differences in time-distance parameters using body weight and height as covariates. It was theorized that dependent variable(s) having significant differences between healthy elders and elders with impairment that are not significantly different between healthy young and healthy elders, would be deemed sensitive to mobility impairments. We hypothesized that power and work calculations would be sensitive to impairments while time-distance measures would not. Because there were 10 dependent variables compared for analysis of kinetic data, an α level of 0.005 was selected to indicate significant differences. Predicted between-groups differences by peak power and by integrated power were then compared.

RESULTS

Age, Body Size, and Time-Distance Parameters

Subject sample averages and standard deviations for age, height, and weight are shown in **Table 1.**

Table 1.

Demographics for the three study groups.

Parameter	Healthy Young N=20 Mean (1 SD)*	Healthy Elders N=16 Mean (1 SD)	Impaired Elders N=24 Mean (1 SD)
Age, yrs	27.00 (4.16)**	72.45 (5.59)	73.52 (7.24)
Height, m	1.65 (0.07)+	1.61 (0.08)	1.59 (0.07)
Weight, kg	61.56 (6.85)+	66.99 (14.69)	71.58 (13.55)

* SD = standard deviation; ** Significantly different from healthy and impaired elders, $p<0.0001$; + Significantly different from impaired elders, $p<0.03$.

The young subjects were taller and lighter ($p<0.03$) than the impaired elders, but not significantly different from the healthy elders. The healthy and impaired elders were not significantly different in age, height, and weight from each other. Time-distance parameters are shown in **Table 2.** Analysis of variance revealed that healthy young women translated their CG significantly faster than both healthy and functionally limited old women ($p<0.0001$) and spent less time in stance phase than both elderly groups ($p<0.005$). CG velocity and stance duration were not significantly different between healthy and impaired elders and there were no significant differences in single support time between healthy young, healthy elderly or impaired elderly groups. When controlling for height and weight, CG velocity remained the only significant between-group difference ($p<0.001$).

Table 2.

Time-distance parameters for the three study groups.

Parameter	Healthy Young N=20 Mean (1 SD)*	Healthy Elders N=16 Mean (1 SD)	Impaired Elders N=24 Mean (1 SD)
CG velocity, m sec^{-1}	1.36 (0.12)**	1.16 (0.21)	1.07 (0.12)
Stance Duration, sec	0.60 (0.03)+	0.64 (0.05)	0.64 (0.03)
Single Support, sec	0.34 (0.02)	0.33 (0.03)	0.32 (0.07)

* SD = standard deviation; ** Significantly different from healthy and impaired elders, $p<0.0001$, and when controlling for height and weight, $p=0.001$; + Significantly different from impaired elders, $p=0.005$; but not significantly different when controlling for height and weight, $p=0.06$.

Joint Power and Work

Net joint muscle power curves for ankle flexion-extension, knee flexion-extension, hip flexion-extension, and hip abduction-adduction are shown in **Figure 2.**

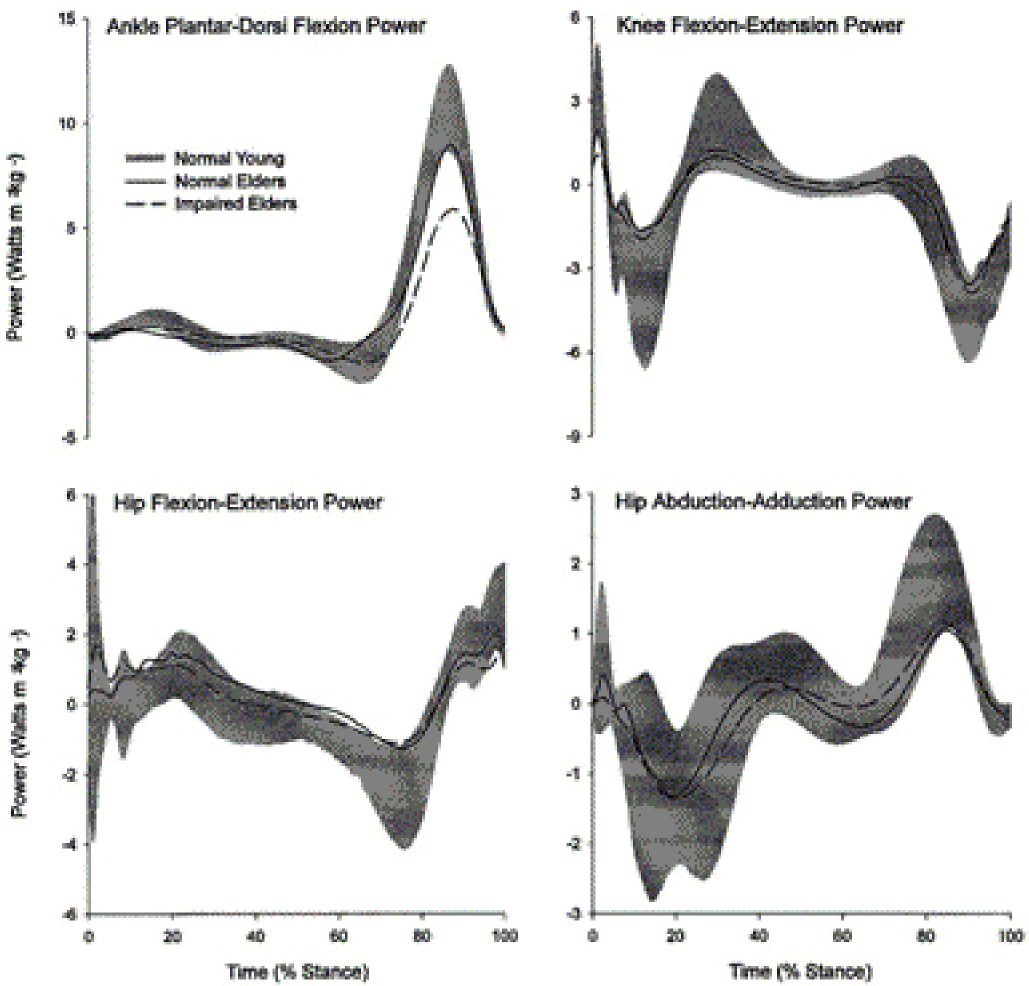

Figure 2.

Power profiles for subjects used in this analysis. The gray shaded area represents the ±1 standard deviation boundary enclosing the young normal mean power. The solid line represents the mean power curves for healthy elderly subjects and the dashed line represents mean power curves for the functionally limited elders. Standard deviations for elders and mean line for young adults are not shown for clarity.

Joint muscle power for healthy young women are depicted by the gray shaded regions, representing ±1 standard deviation from their population mean (N=20). Joint muscle power for healthy elderly women (N=16) and functionally limited elderly women (N=24) are represented by their population mean curves (solid line: healthy elders; dashed line: functionally limited elders). Population averages and standard deviations of peak power and integrated power (work) for the three study groups are summarized in **Table 3.**

Table 3.

Mean peak power and integrated power normalized to body mass index (BMI = weight × height^{-2}) for the three study groups. See **Figure 1** for A1 through HF2 definitions.

	Peak Power, Watts m^2 kg^{-1} (1 SD)*			**Integrated Power, Joules m^2 kg^{-1} (1 SD)**		
	Healthy Young	**Healthy Elders**	**Impaired Elders**	**Healthy Young**	**Healthy Elders**	**Impaired Elders**
A1	-1.68 (0.83)	-1.89 (1.00)	-1.60 (0.61)	-0.209 (0.093)	-0.300 (0.161)	-0.237 (0.109)
A2	10.78 (2.03)	9.32 (3.62)	6.08 (2.06)	0.947 (0.173)	-0.850 (0.343)	-0.529 (0.175)
K1	4.20 (2.54)	-2.07 (1.03)	-2.06 (1.19)	0.274 (0.175)	-0.129 (0.078)	-0.136 (0.081)
K2	2.30 (1.72)	1.28 (0.70)	1.42 (0.80)	0.211 (0.164)	-0.121 (0.081)	-0.159 (0.175)
K3	-5.14 (1.54)	-4.00 (3.21)	-4.06 (1.71)	-0.411 (0.183)	-0.321 (0.279)	-0.338 (0.161)
HS1	1.57 (1.06)	1.93 (0.92)	2.50 (1.86)	0.141 (0.268)	0.268 (0.186)	0.239 (0.170)
HS2	-2.80 (1.42)	-1.43 (1.73)	-1.23 (0.74)	-0.379 (0.208)	-0.164 (0.177)	-0.201 (0.156)
HS3	3.14 (1.45)	2.82 (1.54)	1.85 (1.13)	0.160 (0.076)	-0.145 (0.102)	-0.101 (0.072)
HF1	-2.22 (1.56)	-1.90 (0.97)	-1.95 (1.57)	-0.197 (0.128)	-0.168 (0.095)	-0.209 (0.171)
HF2	2.04 (0.90)	1.20 (0.77)	1.29 (0.66)	0.188 (0.098)	0.108 (0.075)	0.131 (0.079)

* SD = standard deviation

Results of the ANCOVA are summarized in **Tables 4** and **5** for peak power and work, respectively. Without controlling for CG velocity, there were numerous differences in joint peak powers and work between the young subjects and the elderly subjects. There were more significant differences between healthy young subjects and impaired elders than between healthy young subjects and healthy elders and between healthy elders and impaired elders. When controlling for CG velocity, all significant differences disappeared except for ankle "push off" power (A2) between healthy young and impaired elders, and between healthy elders and impaired elders. This effect was consistent for both peak power and work.

Table 4.

Stance phase peak power between-groups differences.

Main Effects	**Healthy Young vs Healthy Elders, *p*-value**		**Healthy Young vs Impaired Elders, *p*-value**		**Healthy Elders vs Impaired Elders, *p*-value**	
	Group	**w/ Covariate (GVEL) ***	**Group**	**w/ Covariate (GVEL)**	**Group**	**w/ Covariate (GVEL)**
A1$^{(-)}$ **						
A2$^{(+)}$			<0.00	<0.00	0.001	0.003
K1$^{(-)}$	0.003	0.030	0.001			
K2$^{(+)}$	0.032		0.031			
K3$^{(-)}$			0.035			
HS1$^{(+)}$						
HS2$^{(-)}$	0.013		<0.00			
HS3$^{(+)}$			0.002		0.027	
HF1$^{(-)}$						
HF2$^{(+)}$	0.006		0.003			

ANOVA main effects with and without CG velocity (GVEL corrected for height) as a covariate. *p*-values >0.05 are not shown (*p*-values <0.005 were considered significant). *Main effect when controlling for GVEL (mean stance phase CG velocity corrected for height). **Superscript (+) denotes generation phase and superscript (-) denotes absorption phase.

Table 5.

Stance phase integrated power (work) between-groups differences.

Main Effects	Healthy Young vs Healthy Elders, *p*-value	Group w/ Covariate (GVEL)*	Healthy Young vs Impaired Elders, *p*-value	Group w/ Covariate (GVEL)	Healthy Elders vs Impaired Elders, *p*-value	Group w/ Covariate (GVEL)
A1$^{(-)}$**	0.041					
A2$^{(+)}$			<0.00	<0.00	<0.00	0.001
K1$^{(-)}$	0.004	0.018	0.001			
K2$^{(+)}$						
K3$^{(-)}$						
HS1$^{(+)}$	0.018	0.025	0.035			
HS2$^{(-)}$	0.002		0.002			
HS3$^{(+)}$			0.012			
HF1$^{(-)}$						
HF2$^{(+)}$	0.011		0.040			

ANOVA main effects with and without CG velocity (GVEL corrected for height) as a covariate. *p*-values >0.05 are not shown (*p*-values <0.005 were considered significant). *Main effect when controlling for GVEL (mean stance phase CG velocity corrected for height). **Superscript (+) denotes generation phase and superscript (-) denotes absorption phase.

DISCUSSION

Our results suggest that decreased ankle plantar-flexor power in late stance phase of gait is a factor associated with functional limitations beyond those of normal aging. In the sample studied here, ankle plantar-flexion power and work were significantly higher for healthy elderly compared to functionally limited elders (p<0.003), and for young compared to functionally limited elders (p<0.0001), but were not significantly different between young and old healthy subjects. This latter finding was surprising, as significant decreases in ankle plantar-flexion power are reported to occur for healthy elderly subjects compared to healthy young subjects (19,26), which was not apparent with our healthy subjects before or after controlling for CG velocity. Although Judge et al. (26) found significantly lower ankle plantar-flexor peak powers during gait for healthy old subjects compared to young subjects (p=0.007), they reported that the difference was not significant when controlling for step length. Winter et al. (19) also found a significant age-related decrease in ankle plantar-flexor power (p<0.01) but did not control for CG velocity or other time-distance parameters, as their study design was aimed more at characterizing changes in kinetic patterns of gait with age. It is also plausible that subjects studied by Winter et al. (19) and Judge et al. (26) had functional limitations, whereas our healthy subjects were more fit. These past studies (19,26) and the results from our study emphasize the importance of controlling for confounding factors when using kinetic data to characterize impairment-related alterations in gait and other locomotor activities.

We found many differences in power and work (summarized in **Tables 3** and **4**) at the knee and hip, but most of these differences disappeared when controlling for CG velocity. Judge et al. (26) reported that elders used more hip flexor power in late stance (HS3) than did younger subjects, but only when correcting for step length. Our study showed that peak power and work of HS3 were greatest for young subjects and least for elders (**Table 3**), consistent also with the findings of Prince et al. (34). Although Winter et al. (19) reported a significant increase in knee work absorption (K3) of healthy elders when compared with young adults ($p<0.01$), we found that peak power and work of K3 were lower for elderly subjects compared with the young subjects, consistent with the findings of Judge et al. (26). Although many possible reasons for such discrepancies may exist, one possible factor is that we had our subjects walk at a predetermined cadence (120 BPM) to eliminate this variable as a possible confounder. Because one's preferred cadence is likely to be determined centrally, impaired elders walking at 'faster than normal' speeds may be suboptimal and affect the resulting power profiles at the joints.

The difficulty in identifying differences in hip and knee power with these subjects may be the effect of subject-to-subject variability. Knee and hip powers are much lower than that of the ankle but with greater variability than the ankle (19), and thus may have obscured any subtle compensations used by this sample of subjects. Because CG velocity did not explain differences in ankle power between healthy young and impaired elders, it might be suggested that the ankle joint acts as a passive-elastic mechanism to aid in advancing the leg into swing phase. If this were the case, however, then the impaired elders would have been expected to compensate with increased hip extensor power in late stance, as reported by Judge et al. (26).

The only dependent variable in our analysis that increased (though not significantly) with age and impairment was hip extensor power in early stance (HS1, **Tables 3** and **4**). Although not conclusive from our data, perhaps the older, weaker subjects used more hip extensor power in early stance to maintain upper body momentum to compensate for weak knee muscles. It might then be speculated that the same subjects also used more contralateral hip extensor power that would correspond to the timing of ipsilateral (the leg analyzed) ankle "push off." This explanation is not conclusive and may require a more detailed analysis, such as segmental and/or muscle power flow (or energy transfer) analysis, to identify the mechanisms of interlimb compensations. We did not investigate the upper body's role in power requirements at the limbs, which may also reveal information that is more conclusive.

Although equivocal, our study suggests that net power and work calculations at the ankle joint are sensitive to functional limitations, while those at the knee and hip appear not to be. We found that time-distance parameters, however, were not sensitive measures of functional limitations in the elderly group studied. Single support time, in particular, was not significantly different across groups ($p=0.34$); however, our ability to only estimate this parameter from the vertical force-plate trace may have influenced the between-group comparisons. Stance duration, a more precise measure, was briefer for healthy young subjects (p=0.005), but not significantly different from elders when controlling for height and weight (p=0.06). Although CG velocity was higher for the young subjects compared with both elderly groups (p<0.001), CG velocity was not significantly different between healthy elders and functionally limited elders (p>0.05). Because the cadence of the subject's gait was controlled, the above result indicates that the elderly subjects took shorter steps. Although useful for providing a general description of gait, time-distance measures cannot be used to identify *how* movements of adjoining segments compensate for mobility impairments. Joint muscle power and work are more closely associated with the physiology of human movement, and, therefore, have the potential to offer more clinically relevant information.

Calculations of net muscle power and work from gait data may be susceptible to misinterpretation when influences, such as (but not limited to) age, weight, stature, and walking velocity are not controlled for. Other factors not addressed here, such as gender (only women were studied), strength, physical fitness (e.g., cardiovascular), and emotional state, which are reported to be related to functional ability (2,3,6), may also influence the power profiles at leg joints during ambulation. Modeling and measurement related factors, such as inertial parameter estimations and joint center errors, can also adversely affect the reliability of kinetic calculations. We have shown previously that linear segmental power calculations (net joint force multiplied by segment endpoint velocity) are sensitive to joint gap velocity discrepancies and, to a smaller degree, numerical differentiation of displacement data (32). Although the latter influences are not trivial, measurement errors are generally random in nature and, therefore, should not skew population comparisons appreciably. Furthermore, the angular power calculations we report here are more stable than linear power calculations (32) and are probably reliable measures of net muscular effort.

To minimize confounding factors in the statistical design, we eliminated body size influence by correcting power and work for body mass index, and our statistical comparison controlled for CG velocity. Our decision to use body mass index (weight over squared height) to normalize power and work calculations, as opposed to the more popular method of normalizing to mass only, warrant justification as follows. Although a higher body mass in young, healthy individuals may be associated with more muscle mass, hence power capacity, the same cannot be said for elders who are heavy but have a much higher concentration of subcutaneous and interstitial muscle fat (35); body mass index is one way to correct for this effect.

CONCLUSIONS

We conclude that decreased ankle plantar-flexor power in late stance-phase of gait may be an important factor in characterizing mobility impairments (time-distance parameters were not sensitive to functional limitations in the group studied) and that joint peak power and integrated power demonstrated similar discriminatory abilities and may be equally useful for analysis of normal and impaired mobility characteristics.

ACKNOWLEDGMENTS

The authors thank Michael S. Puniello, MS, PT and Carol A. Tucker, PhD, PT for assistance in data analysis and insightful comments.

REFERENCES

1. Shumway-Cook A, Gruber W, Baldwin M, Shiquan L. The effect of multidimensional exercises on balance, mobility, and fall risk in community-dwelling older adults. Phys Ther 1997;77:46-57.
2. Wolfson L, Judge J, Whipple R, King M. Strength is a major factor in balance, gait, and occurrence of falls. J Gerontol Med Sci 1995;50A:64-7.
3. Gibbs J, Hughes S, Dunlop D, Singer R, Chang RW. Predictors of change in walking velocity in older adults. J Am Geriatr Soc 1996;44:126-32.
4. Campbell WW, Crim MC, Dallal GE, Young VR, Evans WJ. Increased protein requirements in the elderly: new data and retrospective reassessments. Am J Clin Nutr 1994;60:501-9.
5. Fiatarone MA, O'Neill EF, Ryan ND, et al. Exercise training and nutritional supplementation for physical frailty in very old people. N Engl J Med 1997;330:1769-75.
6. Buchner DM, Cress ME, Esselman PC, et al. Factors associated with changes in gait speed in older adults. J Gerontol Med Sci 1996;51:M297-302.
7. Cwikel J, Fried AV, Galinsky D, Ring H. Gait and activity in the elderly: implications for community falls-prevention and treatment programs. Disabil Rehabil 1995;17:277-80.
8. Guimaraes RM, Isaacs B. Characteristics of the gait in old people who fall. Int Rehabil Med 1980;2:177-80.
9. Hadley E, Radebaugh TS, Suzman R. Falls and gait disorders among the elderly. A challenge for research. Clin Geriatr Med 1985;1:497-500.
10. Wolfson LI, Whipple R, Amerman P, Kaplan J, Kleinberg A. Gait and balance in the elderly. Two functional capacities that link sensory and motor ability to falls. Clin Geriatr Med 1985;1:649-59.
11. Wolfson L, Whipple R, Amerman P, Tobin JN. Gait assessment in the elderly: a gait abnormality rating scale and its relation to falls. J Gerontol Med Sci 1990;45:M12-9.
12. Ferrandez AM, Pailhous J, Durup M. Slowness in elderly gait. Exp Aging Res 1990;16:79-89.
13. Himann JE, Cunningham DA, Rechnitzer PA, Paterson DH. Age-related changes in speed of walking. Med Sci Sports Exerc 1988;20:161-6.
14. Larish DD, Martin PE, Mungiole M. Characteristic patterns of gait in the elderly. Ann NY Acad Sci 1988;515:18-32.
15. Öberg T, Karsznia A, Oberg K. Basic gait parameters: reference data for normal subjects, 10-79 years of age. J Rehabil Res Dev 1993;30:210-23.
16. Ostrosky KM, Van Swearingen JM, Burdett RG, Gee Z. A comparison of gait

characteristics in young and old subjects. Phys Ther 1994;74:637-44.
17. Roach KE, Miles TP. Normal hip and knee active range of motion: the relationship to age. Phys Ther 1991;71:656-65.
18. Wall JC, Hogan DB, Turnbull GI, Fox RA. The kinematics of idiopathic gait disorder. A comparison with healthy young and elderly females. Scand J Rehabil Med 1991;23:159-64.
19. Winter DA, Patla AE, Frank JS, Walt SE. Biomechanical walking pattern changes in the fit and healthy elderly. Phys Ther 1990;70:340-7.
20. Brown M, Sinacore DR, Host HH. The relationship of strength to function in the older adult. J Gerontol Med Sci 1995;50A:55-9.
21. Young A, Skelton DA. Applied physiology of strength and power in old age. Int J Sports Med 1994;15:149-51.
22. Shultz AB. Muscle function and mobility biomechanics in the elderly: an overview of some recent research. J Gerontol 1995;50A:60-3.
23. Bassey EJ, Fiatarone MA, O'Neill EF, Kelly M, Evans WJ, Lipsitz LA. Leg extensor power and functional performance in very old men and women. Clin Sci 1992;82:321-7.
24. Skelton DA, Greig CA, Davies JM, Young A. Strength, power and related functional ability of healthy people aged 65-89 years. Age Ageing 1994;23:371-7.
25. Rantanen T, Avela J. Leg extension power and walking speed in very old people living independently. J Gerontol Med Sci 1997;52A:M225-31.
26. Judge JO, Davis RB, Ounpuu S. Step length reductions in advanced age: the role of ankle and hip kinetics. J Gerontol Med Sci 1996;51A:M303-12.
27. Bendall MJ, Bassey EJ, Pearson MB. Factors affecting walking speed of elderly people. Age Ageing 1989;18:327-32.
28. Jette AM, Lachman M, Giorgetti MM, et al. Exercise: it's never too late! Am J Pub Health, In press.
29. Young, Chandler, Snow, Robinetto, Zehner, Loftberg. Anthropometric and mass distribution characteristics of the adult female. Report No. FAA-AM-83-16 (Government accession No. AD-A143096), Wright-Patterson Air Force Base, OH, 1983.
30. Tupling SJ, Pierrynowsky MR. Use of cardan angles to locate rigid bodies in three-dimensional space. Med Biol Eng Comput 1987;25:527-32.
31. Jacobs NA, Skorecki J, Charnley J. Analysis of the vertical component of force in normal and pathological gait. J Biomech 1972;5:11-34.
32. McGibbon CA, Krebs DE. The influence of segment endpoint kinematics on segmental power calculations. Gait Posture 1998;7:237-42.
33. Eng JJ, Winter DA. Kinetic analysis of the lower limbs during walking: what information can be gained from a three-dimensional model? J Biomech 1995;28:753-8.
34. Prince F, Corriveau H, Hebert R, Winter DA. Gait in the elderly. Gait Posture 1997;5:128-35.
35. Reed RL, Pearlmutter L, Yochum K, Meredith KE, Mooradian AD. The relationship between muscle mass and muscle strength in the elderly. J Am Geriatr Soc 1991;39:555-61.

Contents

Back to Top

---

[1]Functional limitation was based on health status influencing subject response as either "limited or a little" or "limited a lot" to one or more of the following: a) moderate activities, such as moving a table, pushing a vacuum cleaner, bowling, or playing golf; b) lifting or carrying groceries; c) climbing several flights of stairs; d) climbing one flight of stairs; e) bending, kneeling, or stooping; f) walking more than a mile; g) walking several blocks; h) walking one block; and i) bathing or dressing one's self. (Return to text.)