

Propulsive adaptation to changing gait speed

Patrick O. Riley^{a,*}, Ugo Della Croce^{a,b}, D. Casey Kerrigan^a

^a*Spaulding CRS Rehabilitation Engineering Laboratory and Hospital, and Department of PM&R Harvard Medical School, 125, Nashua Street, Boston, MA 02114, USA*

^b*Dipartimento Scienze Biomediche, Universita' di Sassari, Italy*

Accepted 20 August 2000

Abstract

Understanding propulsion and adaptation to speed requirements is important in determining appropriate therapies for gait disorders. We hypothesize that adaptations for changing speed requirements occur primarily at the hip. The slow, normal and fast gait of 24 healthy young subjects was analyzed. The linear power was analyzed at the hip joint. The anterior–posterior and vertical induced accelerations of the hip were also determined. Linear power and anterior–posterior-induced acceleration (IA) analyses of the hip reveal that the lower limb joint's moments contribute to body forward propulsion primarily during late swing and early stance. Propulsive adaptations to speed changes occur primarily at the hip and secondarily at the ankle. These analyses show that hip muscles, particularly the hip extensors, are critical to propulsion. They also show that ankle function is primarily for support, but is important to propulsion, especially at slow speeds. © 2001 Elsevier Science Ltd. All rights reserved.

Keywords: Gait analysis; Biomechanics; Power; Induced accelerations; Speed

1. Introduction

Clinicians seeking to rehabilitate patients with gait disorders must take into account a range of factors affecting patient safety, comfort, strength reserve and endurance. It is essential to provide the patient with adequate support and propulsion using the strength and control available to the patient. Joint powers, measured by gait analysis, are used to guide this aspect of rehabilitation (Winter, 1991). As a result, there has been a strong emphasis on developing ankle plantarflexor “push-off” power (Winter, 1983; Olney et al., 1990, 1991; Ounpuu et al., 1996). Winter (1983) reported that, in healthy adults, ankle joint power decreased with decreasing speed. However, Chen et al. (1997) reported that, in healthy children, ankle joint power was nearly constant and contributed relatively less to propulsion as speed increased. Moreover, joint powers indicate the net power developed by muscles acting about a joint; they do not provide information as to the function of that power (Meinders et al., 1998).

Several model-based techniques are available to elucidate function, to determine the contribution of power

developed at one joint to the kinematics and kinetics of all parts of the body. Buczek et al. (1994) suggested the usefulness of linear power analysis and Meinders et al. (1998) skillfully used a combination of joint power, linear power and segmental energy flow analyses to examine the function of ankle “push-off” power. Induced acceleration analysis has been used to study posture control (Zajac and Gordon, 1989) and gait (Meglan, 1991; Kepple et al., 1997; Riley and Kerrigan, 1999). A combination of linear power and induced acceleration analyses measured subtle differences in propulsion between healthy elders and young control subjects (Riley et al., in review).

The latter study led us to conclude that linear power and induced acceleration analyses provided a profound insight into the biomechanics of propulsion. However, this work also brought out the necessity of determining how propulsive mechanisms vary with speed, which is the purpose of the present work. Our hypothesis is that adaptation to changing speed requirements occurs primarily by modulating the action of the hip flexors and extensors.

2. Methods

Healthy young volunteers (16 female/8 male) between 18 and 40 years of age (23.9 ± 4.4 years, mean \pm

* Corresponding author. Tel.: + 617-573-2731; fax: + 617-573-2769.
E-mail address: priley@partners.org (P.O. Riley).

standard deviation (S.D.)) were recruited. The subjects had a mean mass of 61.5 ± 12.2 kg and a mean height of 1.68 ± 0.10 m. Each subject provided informed consent before testing in accordance with a protocol approved by the hospital institutional review board. Subjects walked barefoot at their self-selected normal speed on a 10 m walkway. The subjects were then asked to walk “faster than you would normally walk.” Finally they were asked to walk “slower than you normally walk. For each subject left and right lower limb data from three trials at each speed were analyzed.

Gait kinematic and kinetic data were acquired using a six-camera VICON 512 system (Oxford Metrics, Oxford, England) and two AMTI forceplates (AMTI, Newton, MA). Subjects wore retro-reflective markers on the left and right anterior–superior–iliac spines, lateral femoral condyles, lateral maleoli and fifth metatarsal heads. Markers were also placed on the sacrum, between the left and right posterior–superior–iliac spines and on lateral wands at the left and right mid-femur and mid-shank. The data were processed using VICON Clinical Manager. The resulting hip, knee and ankle joint center positions, hip and knee rotation angles, and pelvis and foot orientation angles provided the input data for model-based analysis.

SIMM/Dynamics Pipeline (Musculographics Inc, Evanston, IL) and SD/Fast (Symbolic Dynamics, Moun-

tainview, CA) were then used to analyze the linear power and induced accelerations (IAs) in a three-step process. First, best-fit model kinematics were estimated from the VICON joint position data. Second, a conventional inverse dynamics analysis was performed using the model kinematics and forceplate data. Third, the generalized forces defined by the inverse dynamic analysis were applied individually to the model on a frame-by-frame basis to determine their contribution to the joint linear powers and model degree of freedom IAs.

Linear power was defined as the dot product of the net joint reaction force and the joint linear velocity. Linear power is a measure of the power flow across a joint, while joint power is a measure of the power generated at the joint. Linear powers were normalized by body mass while joint powers were normalized by body mass and height because the generalized force in joint powers is a moment. IAs were calculated as described by Kepple et al. (1997) with the foot position constrained at the center of pressure. IAs are a measure of each generalized force’s contribution to the total acceleration of a degree of freedom. The linear powers and IAs at the hip joint were examined in detail.

Unpaired *t*-tests were used to confirm that parameters did not differ between the sexes prior to combining male and female data. Repeated measures analyses of variance were used to determine if linear power and IA parameters

Table 1

Gait parameters, combined male and female, three trials, left and right side average (S.D.). $p < 0.001$ except where noted

	Slow	Normal	Fast
<i>Time distance parameters</i>			
Velocity (m/s)	0.87 (0.12)	1.19 (0.13)	1.74 (0.21)
Cadence (1/s)	0.78 (0.08)	0.94 (0.06)	1.16 (0.11)
Stride length (m)	0.57 (0.04)	0.63 (0.06)	0.76 (0.07)
<i>Kinematic parameters</i>			
Max hip flexion (°)	30.2 (6.3)	31.5 (5.8)	36.4 (6.0)
Max hip extension (°)	7.9 (5.7)	9.7 (6.5)	13.1 (6.7)
Max knee flexion (°)	54.7 (4.5)	57.6 (3.7)	60.0 (4.4)
Max ankle dorsiflexion (°)	12.3 (2.9)	11.3 (3.2) ^a	9.4 (3.0)
Max ankle plantarflexion (°)	15.2 (4.6)	17.7 (4.6) ^b	19.9 (4.2)
<i>Kinetic parameters (normalized to body mass and height)</i>			
Hip flex moment max (N m/kg m)	0.31 (0.10)	0.47 (0.14)	0.78 (0.23)
Hip ext moment max (N m/kg m)	0.30 (0.08)	0.40 (0.09)	0.62 (0.14)
Knee flex moment max (N m/kg m)	0.10 (0.08)	0.19 (0.12)	0.44 (0.17)
Knee ext moment max (N m/kg m)	0.28 (0.08)	0.31 (0.08) ^b	0.33 (0.08) ^b
Ankle plantarflex moment max (N m/kg m)	0.86 (0.06)	0.92 (0.07)	0.99 (0.07)
Ankle dorsiflex moment max (N m/kg m)	0.05 (0.02)	0.07 (0.02)	0.11 (0.04)
Hip jnt. power gen. max (W/kg m)	0.52 (0.14)	0.94 (0.24)	1.99 (0.48)
Hip jnt. power abs. max (W/kg m)	0.20 (0.07)	0.33 (0.12)	0.71 (0.24)
Knee jnt. power gen. max (W/kg m)	0.34 (0.12)	0.48 (0.16)	0.85 (0.23)
Knee jnt. power abs. max (W/kg m)	0.35 (0.08)	0.65 (0.13)	1.38 (0.34)
Ankle jnt. power gen. max (W/kg m)	1.26 (0.41)	1.93 (0.54)	2.93 (0.67)
Ankle jnt. power abs. max (W/kg m)	0.56 (0.14)	0.63 (0.12) ^b	0.58 (0.13) ^b

^a $p = 0.001$.

^b $p > 0.001$.

varied with speed. Pearson correlations were used to determine the direction of variation. The significance of changes between slow and normal and between normal and fast speeds was tested using paired *t*-tests. To correct for multiple parameters $p < 0.001$ was required for significance.

3. Results

The self-selected slow (0.87 ± 0.12 m/s), normal (1.19 ± 0.13 m/s), and fast (1.74 ± 0.21 m/s) speed gaits of healthy young adults were analyzed. Normal speed gait temporal, kinematic and kinetic parameters were within normal limits (Table 1). There were no significant differences between males and females in any of the standard gait parameters or in the linear power and IA parameters.

Linear power analyses indicated that each limb provided power to the upper-body in late swing, early stance and just before toe-off (Fig. 1). In mid to late stance and early swing, the upper-body provided linear power to the lower limb. In late stance positive power was produced by the ankle moments, consistent with the concept of “push-off.” As speed increased, the late swing and early stance positive power peaks increased in magnitude, as did the mid stance and early swing negative power peaks. However, the magnitude of the “push-off” linear power peak decreased slightly with increasing speed (slow 0.6 ± 0.3 W/kg; fast -0.2 ± 0.7 W/kg, $p < 0.001$). The late stance positive ankle moment linear power contribution increased significantly (slow 1.4 ± 0.4 W/kg; normal 2.1 ± 0.6 W/kg, $p < 0.001$; fast 3.2 ± 0.8 W/kg, $p < 0.001$), but this was more than offset by negative contributions from the knee (slow -0.14 ± 0.05 W/kg; normal -0.27 ± 0.11 W/kg, $p < 0.001$; fast -0.70 ± 0.40 W/kg, $p < 0.001$) and hip moments (slow -0.5 ± 0.2 W/kg; normal -0.9 ± 0.2 W/kg, $p < 0.001$; fast -2.2 ± 0.7 W/kg, $p < 0.001$).

Linear power is a scalar parameter combining anterior–posterior (AP), vertical and medial–lateral kinetics. The AP and vertical IAs of the hip joint provided explicit vector measures of propulsive and support function of each generalized force (constraint force or joint moment). For these analyses, the contributions of all ipsilateral leg moments were combined to estimate the IA due to that limb (IA-leg, Fig. 2A and B). Similarly, the contributions of the horizontal, vertical and lateral pelvic constraint forces, as well as the pelvic flexion/extension, rotation and obliquity constraint moments were combined. This combined parameter measured the contribution of the contralateral limb and pelvic rotations (IA-cleg, Fig. 2A and B). The propulsive contributions of the two limbs were approximately mirror images of each other, but phase shifted in proportion to double support time (Fig. 2A). Each limb alternately provided support (a

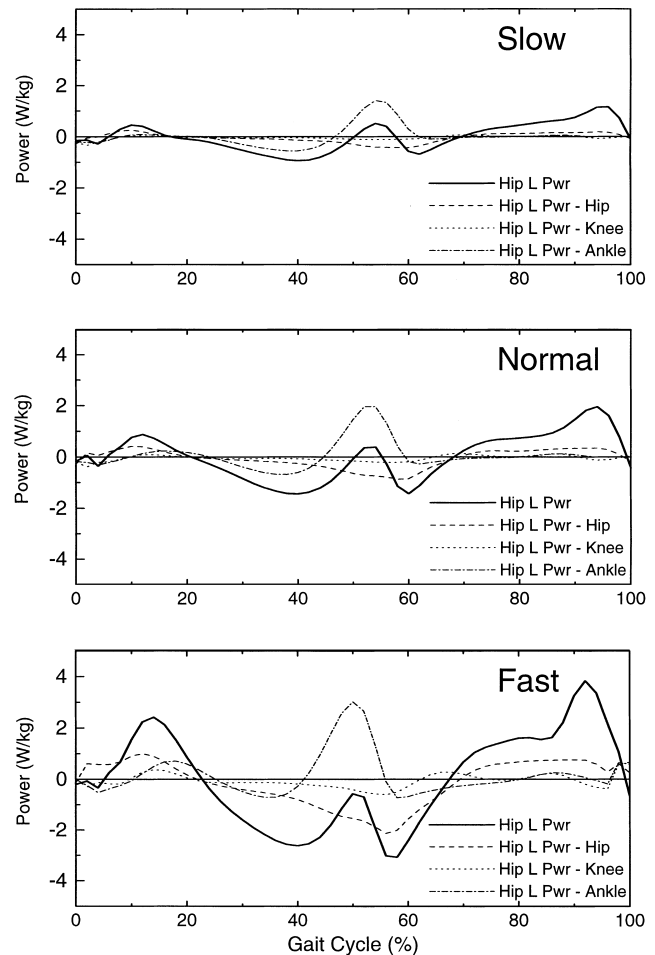


Fig. 1. Hip linear power, normalized to body mass, at slow, normal and fast speeds. Hip L PWR — the total linear power at the hip. Positive indicates transfer from the leg to the upper-body. Hip L PWR-Hip, Hip L PWR-Knee, and Hip L PWR-Ankle — the respective contributions of the hip, knee and ankle net moments to total hip linear power. Normalize to one gait cycle; heel strike to heel strike.

positive vertical IA) to balance the negative vertical IA due to gravity and inertial terms (IA-cg) (Fig. 2B). The sum of all IAs (IA-Tot) corresponded closely to the total kinematic acceleration (Acc) except in very early stance. Hip AP IA analysis indicated that the late swing and early stance positive linear power peaks corresponded to periods when the ipsilateral limb was making a positive contribution to forward propulsion, i.e. IA-leg was positive (Fig. 2A). However, the ipsilateral limb retarded forward propulsion from mid stance through early swing and, thus, did not contribute to the “push-off” positive linear power. The contralateral limb maintained forward motion through late stance and early swing. Late stance positive linear power was due to the ipsilateral limb’s late stance support contribution (IA-cleg, Fig. 2B).

The overall pattern of the ipsilateral limb’s AP contribution (propulsive in late swing and early stance, retarding in late stance and early swing) remained constant as

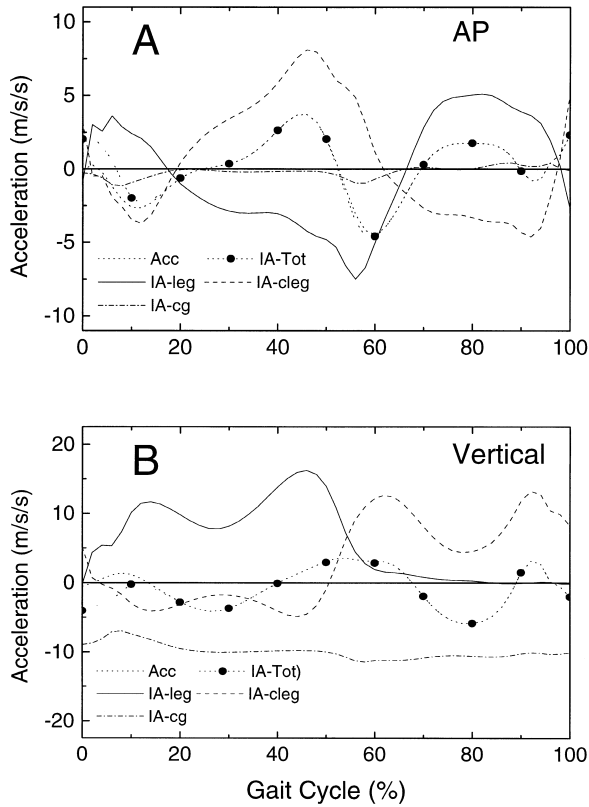


Fig. 2. The AP (A) and vertical (B) induced linear accelerations of the hip joint at fast speed. Positive values indicate accelerations forward and up. Acc — measure acceleration. IA-Tot — sum of the induced accelerations. IA-leg — induced accelerations due to the ipsilateral leg moments. IA-cleg — induced accelerations due to moments on the contralateral leg and pelvis. IA-cg — induced accelerations due to gravity and inertia.

speed varied. However, the magnitude of the contribution changed with speed, as did the relative contribution of the hip and ankle joint moments (Fig. 3). The contribution of knee moment to propulsion was minimal at all speeds. The hip extensor moment accounted for the positive IAs in late swing and early stance. The hip flexor moment produced the retarding IA in late stance and early swing. In late stance, the latter was opposed by a large positive contribution due to the ankle plantarflexors. As speed increased, the ankle component increased, but was relatively constant (slow 3.7 ± 0.7 m/s/s; fast 5.6 ± 1.3 m/s/s, $p < 0.001$). The dominant hip contribution becomes significantly more negative as speed increases (slow -5.3 ± 1.1 m/s/s; norm -6.7 ± 1.2 m/s/s, $p < 0.001$; fast -10.5 ± 2.1 m/s/s, $p < 0.001$).

The overall pattern and magnitude of each limb's contribution to support remained relatively constant at different speeds. (Fig. 4). The peak ankle contribution increased only slightly with speed (slow 14.3 ± 0.7 m/s/s; normal 14.9 ± 0.8 m/s/s, $p < 0.001$; fast 15.5 ± 1.0 m/s/s, $p < 0.001$). The primary vertical load, body weight, re-

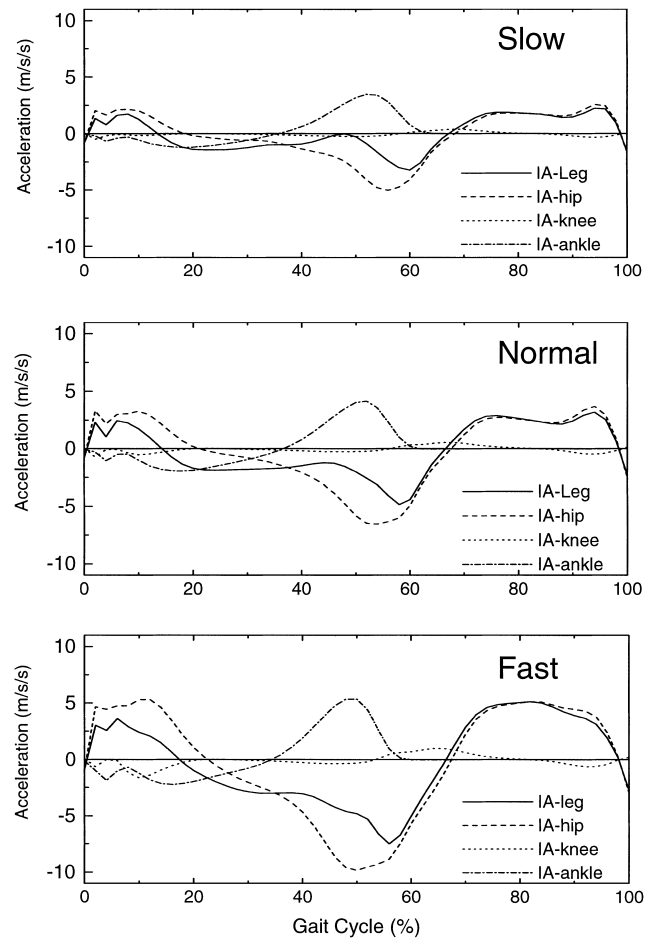


Fig. 3. Joint moment contributions to hip AP accelerations. IA-leg — induced accelerations due to the ipsilateral leg moments. IA-hip, IA-knee, and IA-ankle — induced accelerations due, respectively, to the hip, knee and ankle moment.

mains constant, only the inertial terms change with speed. The ankle moments accounted for essentially all of the leg's support contribution, implying that the ankle, together with pelvic rotations, is primarily responsible for controlling the vertical excursion of the body's center of gravity.

4. Discussion

The normal gait temporal, kinematic and classic kinetic parameters correspond well to that reported in the literature (Murray et al., 1964, 1966, 1970; Kadaba et al., 1989, 1990; Kerrigan et al., 1998). The variations in these parameters with speed agree with those reported by Winter (1983). The overall power analysis of normal speed gait agrees with that reported by Meinders et al. (1998). This study supports his conclusion that ankle plantarflexor power has a limited role in propelling the body in

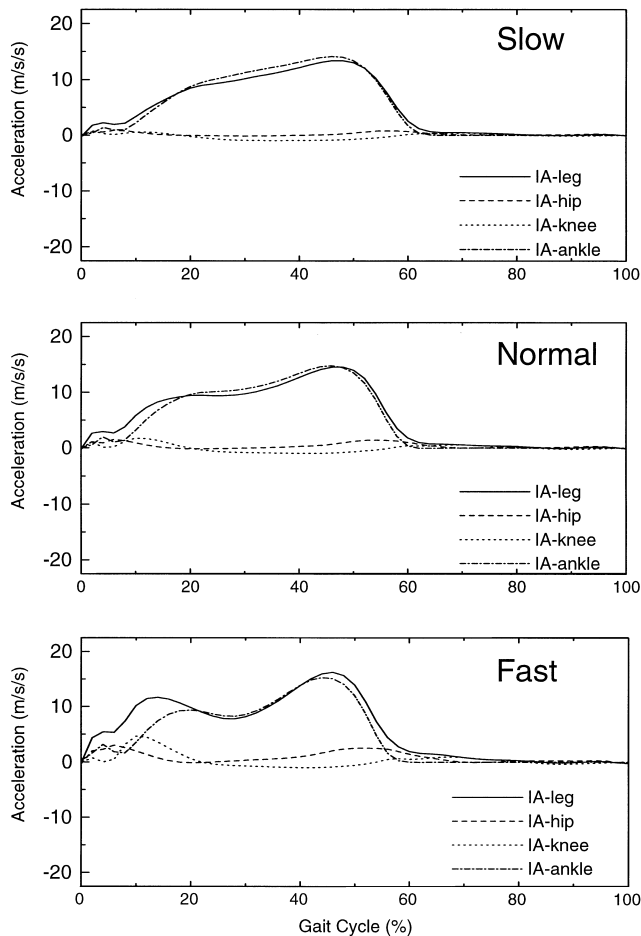


Fig. 4. Joint moment contributions to hip vertical accelerations. IA-leg — induced accelerations due to the ipsilateral leg moments. IA-hip, IA-knee, and IA-ankle — induced accelerations due, respectively, to the hip, knee and ankle moment.

late stance. Using segment energy analysis, he showed that ankle plantarflexor power provides kinetic and potential energy to the leg, which is recovered for propulsive effect, in late swing. The current analyses of overall linear power and the contributions of the various joint moments substantially agree with analyses performed in this laboratory on a similar but smaller population of healthy young subjects who were compared to healthy elders (Riley et al., in review). The latter analyses were performed on data derived from a different motion analysis system and using a slightly different limb model.

The result that the stance leg was propulsive in early stance and retarded upper-body forward motion in late stance is counter-intuitive. The ground reaction force (GRF) is directed posteriorly in early stance and forward before toe-off. Thus, the stance limb must be retarding forward motion at initial contact and propelling the body forward before toe-off. However, the GRF acts on the whole body and is shaped by the body mass and inertia,

as well as the joint moments. In this analysis all gravity and inertial effects, including those of each leg as well as the upper-body, are incorporated in a single term, IA-cg (Fig 2A). The leg terms (IA-leg and IA-cleg) account exclusively for the actions of joint moments and the muscles that produce them. Thus, while as a result of body kinematics the GRF may be acting through the leg to retard forward motion, the muscles in the leg may be acting to propel the body forward. It is the latter effect that is being measured by the IA-leg and IA-cleg parameters.

While these results alter the classic view of ankle function, they also highlight the relative importance of the ankle moments. The ankle moment was responsible for support throughout stance. In late stance, the ankle plantarflexor moment required for support generated a propulsive contribution that was always significant and relatively more so at slow-speeds. Patients in rehabilitation for gait related disabilities typically ambulate at slow speeds, frequently even slower than the slow-speed data. Thus, loss of ankle function impairs support and alters propulsion, requiring compensations at other joints.

The physiological interpretation of these results is relatively straightforward. The leg acts directly to propel the upper-body forward when the hip extensors are acting. In healthy young subjects, adaptation to changing speed requirements occurred primarily in the activation of hip flexors and extensors. The contribution of the ankle is significant, but relatively constant, across the range of walking speeds, dictated primarily by the ankle's role in supporting the upper-body mass. Thus, maintaining or rehabilitating ambulatory function requires maintenance of adequate hip strength and range of motion, and the ability to modulate the activation of muscles acting about the hip.

Acknowledgements

We wish to acknowledge the support of the Shelby Cullom Davis Foundation, the Ellison Foundation and NIH HD01071-05.

References

- Buczek, F.L., Kepple, T.M., et al., 1994. Translational and rotational joint power terms in a six degree-of-freedom model of the normal ankle complex. *Journal of Biomechanics* 27 (12), 1447–1457.
- Chen, I.H., Kuo, K.N., et al., 1997. The influence of walking speed on mechanical joint power during gait. *Gait and Posture* 6 (3), 171–176.
- Kadaba, M.P., Ramakrishnan, H.K., et al., 1989. Repeatability of kinematic, kinetic, and electromyographic data in normal adult gait. *Journal of Orthopaedic Research* 7 (6), 849–860.
- Kadaba, M.P., Ramakrishnan, H.K., et al., 1990. Measurement of lower extremity kinematics during level walking. *Journal of Orthopaedic Research* 8 (3), 383–392.

- Kepple, T.M., Siegel, K.L., et al., 1997. Relative contributions of the lower extremity joint moments to forward progression and support during gait. *Gait and Posture* 6 (1), 1–8.
- Kerrigan, D.C., Todd, M.K., et al., 1998. Gender differences in joint biomechanics during walking: normative study in young adults. *American Journal of Physical Medicine & Rehabilitation* 77 (1), 2–7.
- Meglan, D. A., 1991. Enhanced analysis of human locomotion. PhD Thesis, The Ohio State University, Columbus, OH.
- Meinders, M., Gitter, A., et al., 1998. The role of ankle plantar flexor muscle work during walking. *Scandinavian Journal of Rehabilitation Medicine* 30 (1), 39–46.
- Murray, M.P., Drought, A.B., et al., 1964. Walking patterns of normal men. *Journal of Bone and Joint Surgery — American Volume* 46A (2), 335–360.
- Murray, M.P., Kory, R.C., et al., 1966. Comparison of free and fast speed walking patterns of normal men. *American Journal of Physical Medicine* 45 (1), 8–23.
- Murray, M.P., Kory, R.C., et al., 1970. Walking patterns of normal women. *Archives of Physical Medicine & Rehabilitation* 51 (11), 637–650.
- Olney, S.J., Griffin, M.P., et al., 1991. Work and power in gait of stroke patients. *Archives of Physical Medicine & Rehabilitation* 72 (5), 309–314.
- Olney, S.J., MacPhail, H.E., et al., 1990. Work and power in hemiplegic cerebral palsy gait. *Physical Therapy* 70 (7), 431–438.
- Ounpuu, S., Bell, K.J., et al., 1996. An evaluation of the posterior leaf spring orthosis using joint kinematics and kinetics. *Journal of Pediatric Orthopedics* 16 (3), 378–384.
- Riley, P. O., Della Croce, U., et al., Effect of Advanced Age on Body Propulsion. in review.
- Riley, P.O., Kerrigan, D.C., 1999. Kinetics of stiff-legged gait: induced acceleration analysis. *IEEE Transactions on Rehabilitation Engineering* 7 (4), 420–426.
- Winter, D.A., 1983. Energy generation and absorption at the ankle and knee during fast, natural, and slow cadences. *Clinical Orthopaedics and Related Research* 175, 147–154.
- Winter, D.A., 1991. The biomechanics and motor control of human gait: normal, elderly and pathological. University of Waterloo Press, Waterloo, Ont.
- Zajac, F.E., Gordon, M.S., 1989. Determining muscle's force and action in multi-articular movement. *Exercise and Sport Sciences Review* 17, 187–230.