Biomechanics of Walking and Running

Philip E. Martin, Arizona State University
David J. Sanderson, University of British Columbia - Vancouver

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CHAPTER 42

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Walking and running are two highly complex motor skills that incorporate input from multiple levels of the nervous system, involve muscular contributions throughout the body, and require the coordination of many skeletal degrees of freedom. Nevertheless, under nonpathologic conditions, an individual can produce a highly stable and repeatable gait pattern without conscious control of the process.

A substantial body of research literature on the biomechanical aspects of walking and running already exists and is rapidly growing. A simple search of the computer-based Medical Literature Analysis and Retrieval System (MEDLINE) of the United States National Library of Medicine yields more than 1000 references written since 1966 for both biomechanics of walking and biomechanics of running. Given the extensive and diverse literature base and the complexity of the motor skills, developing a comprehensive review of the biomechanics of walking and running in a single chapter is a daunting, if not impossible, task.

There already exist many published review articles and books that focus broadly on the biomechanical nature of walking and/or running (e.g., 1–14) as well as many more focused reviews on the subject (e.g., 15–22). Therefore, we have not attempted to produce a comprehensive review of walking and running biomechanics in this chapter, but rather we have chosen a small subset of topics that we feel are interesting and timely and that complement existing reviews. These include the relationships between gait biomechanics and economy, age-related changes in walking mechanics, amputee locomotion, the role of elastic mechanisms in gait, and the adaptability of gait. The reader is encouraged to seek other original and review sources for alternative perspectives on gait biomechanics.

RELATIONSHIPS BETWEEN GAIT BIOMECHANICS AND ECONOMY

Considerable research has focused on the aerobic demand and energy cost of walking and running, and their associations with biomechanical features of gait. Economy, defined as the steady-state aerobic demand for a given submaximal task, has performance implications for the runner seeking to maximize distance running performance as well as for special clinical populations that have heightened energy cost demands and relatively low physical working capacities. Research has demonstrated that gait economy tends to vary widely among individuals, and that this variation exists in the absence of neurologic and musculoskeletal deficiencies and disease that can increase energy cost substantially. As an example, Daniels and colleagues (23) reported aerobic demands for 13 competitive distance runners ranging from 39.5 to 45.1 mL/kg/min for graded walking at 1.78 m/s, 43.2 to 53.8 mL/kg/min for level running at 4.13 m/s, and 33.2 to 43.0 mL/kg/min for a 30-cm step test. Interestingly, Daniels and colleagues also observed that the economy measures for different forms of exercise were not strongly associated with one another. They suggested that economy “is not a function of inherent muscular metabolic economy” (p. 613) but rather may be related to either skill level or anatomic differences that contribute to different muscle mechanical advantages about joints. The question then is whether selected biomechanical and/or structural factors explain economy differences between individuals.

Gait Speed

The speed at which one locomotes is perhaps the simplest and most fundamental descriptor of gait kinemat-
ics. Altering gait speed is also one of the most common ways of manipulating exercise intensity, in both research and recreational exercise settings. Aerobic demands expressed per unit of time (mL/kg/min) for both walking and running increase systematically as speed, and thus exercise intensity, increases (Fig. 42–1). When the aerobic demand or energy cost is considered relative to the distance traversed (e.g., mL/kg/km), it becomes apparent that walking and running economy responses are substantially different from one another (Fig. 42–2). The mass- or weight-specific aerobic demand to cover a given distance varies little with running speed, indicating that the total energy cost to run a given distance is nearly the same whether a leisurely pace or a high speed is used. It should be noted that the data for running in Fig. 42–2 are for treadmill locomotion for which wind resistance effects are negligible. Aerobic demand for overground running at 5 m/s is 5% to 7% higher than that for treadmill running (24). Because wind resistance is proportional to speed squared (25), the tendency for the aerobic demand to decline subtly with increasing running speed, as shown in Fig. 42–2, would be diminished for overground running. Thus, it is reasonable to conclude that there is no clearly defined most economical speed of running. In contrast, there is considerable variation in walking aerobic demand per unit of distance traveled as speed changes, and a readily apparent most economical speed of walking (1.3–1.4 m/s) emerges (26,27). Martin and colleagues (26) have shown that this most economical walking speed is quite insensitive to the age and physical activity status of the subjects. Young and elderly, active and sedentary adults had very similar most economical speeds.

In a cross-species assessment of the energy cost of running, Kram and Taylor (28) noted that the energy consumed to move a unit of body weight or mass a given distance (i.e., cost of transport, e.g., J/N/m) changes little with running speed. The running aerobic demand data for humans reflected in Fig. 42–2 fit well with this observation. Kram and Taylor further proposed that the cost of transport for running is inversely proportional to support distance (i.e., the distance the body moves forward during single limb support), and they showed that support distances changed little across a range of running speeds for multiple species. Running support distances for humans, however, increase as running speed increases (29,30), and thus they are not consistent with Kram and Taylor’s model.

The absence of a speed-independent cost of transport for walking (Fig. 42–2) appears to be associated with two factors. At the slow end of the walking speed continuum (i.e., <1.3 m/s), the relative contribution of the cost of resting or maintenance metabolism to the total energy cost increases as speed decreases (26). This results in a relatively high cost of transport at very slow walking speeds despite a low exercise intensity. As walking speed increases, both stride length and rate increase. At the fast end of the walking speed continuum (>1.4 m/s), lower-extremity musculature must not only produce higher forces to generate longer stride lengths but also must produce those forces at higher speeds as stride rate rises. Hill (31) demonstrated that muscle efficiency is a function of shortening velocity such that a most efficient velocity exists. Thus, fundamental force-velocity and associated power-velocity properties of muscle likely contribute to less efficient production of force at very high walking speeds (26).

**Stride Rate/Length**

The aerobic demand of walking and running at any given speed is a function of stride rate (SR) and length (SL), such that aerobic demand increases curvilinearly as SR is either increased or decreased (and thus, SL is either shortened or lengthened, respectively) from the preferred SR (Fig. 42–3) (32–37). An individual’s preferred SR/SL combination usually is in close agreement

![Graph showing aerobic demand vs. speed](image1)

**Fig. 42–1.** The aerobic demands (mL/kg/min) for walking and running increase systematically as speed increases. A curvilinear relationship between aerobic demand and speed is apparent for walking but not for running.

![Graph showing aerobic demands of walking and running](image2)

**Fig. 42–2.** The aerobic demands of walking and running reflect substantially different responses to increases in speed when expressed per unit of distance traveled (mL/kg/km). A most economical speed of walking is apparent for walking but not for running.
with his or her optimal or most economical SR/SL. Morgan and colleagues (37) observed that only 20% of a sample of 45 recreational runners used a preferred SR/SL combination that deviated by 5% or more from their most economical SR/SL. The specific mechanisms underlying the U-shaped SR/SL—aerobic demand response is again linked to fundamental muscle force and power-generating capabilities. Changes in SR and SL require concomitant changes in the rates of muscle lengthening and shortening and rate of force development that ultimately affect aerobic demand (28,38).

Holt and colleagues (39) suggest that stride rate rather than stride length is the critical factor that determines the muscular effort required to generate the gait cycle, and that stride rate is strongly influenced by the inertia characteristics of the swinging limbs. They propose that walking can be modeled as a force-driven harmonic oscillator (FDHO) and that the resonant frequency of the FDHO model, which is dependent on the anthropometric and inertia properties of the legs, corresponds to the preferred rate of walking. Their results supported their hypothesis that “the resonant frequency of a harmonic oscillator can accurately predict that chosen by subjects when appropriate adjustments are made to the formula based on an optimization criterion of minimum force” (p. 64). They concluded that stride rate or stride time is effectively a motor control parameter that is determined by the physical attributes of the system, namely its inertia characteristics. In subsequent research, Holt and colleagues (40) confirmed that preferred stride rate was not different from that predicted from their FDHO model. When stride rate was manipulated during constant speed treadmill walking, both preferred and FDHO-predicted stride rates resulted in minimal aerobic demand, lending additional support to the association between SR/SL and gait economy. Recent research examining preferred and FDHO predicted stride rates for backward walking (41) and for children (42) lends some support for the generalizability of the notion that the body tends to self-optimize gait such that muscular effort and energy cost are minimized. This phenomenon needs to be examined further for other cyclic, submaximal tasks.

**Ground Reaction Forces**

Although various features of the ground reaction force (GRF) have been studied extensively, the association between GRF characteristics and gait economy is not well documented. In their cross-species research involving mammals ranging in mass from 30 g to 140 kg, Kram and Taylor (28) confirmed their hypothesis that the rate of energy consumption per unit of body weight is inversely proportional to the weight-specific rate of force application to the ground, and ultimately to the time of force application, during each stride. Thus, they suggested that simple knowledge of single limb stance time and body weight allows for a good approximation of an animal’s rate of energy consumption during running. In contrast with Kram and Taylor’s hypothesis, Williams and Cavanagh (43,44) found a direct relationship between support time and running aerobic demand in humans \((r=0.49)\), indicating that longer support times were associated with higher rates of energy consumption. While Kram and Taylor’s principle provides a powerful first approximation of the energy cost of running in humans and animals, the utility of this principle for explaining interindividual differences in running economy in humans is unsubstantiated. Williams and Cavanagh (43) also noted that more economical runners are more likely to display lower first peaks in the vertical GRF, smaller anteroposterior and vertical peak forces, and a rear-foot striking pattern. They suggested forefoot strikers place greater demands on the musculature, particularly the plantar flexors, to assist with impact absorption in early stance, whereas rear-foot strikers rely more heavily on footwear and skeletal structures to cushion and support the body in early stance.

**Mechanical Power**

Aerobic demand represents a global descriptor of the physiologic demand of walking or running, whereas mechanical power output reflects a global expression of muscular effort or output. Assuming that a substantial portion of the metabolic demands of walking and running is attributed to muscles doing work (i.e., actively shortening or lengthening), then mechanical power should be an effective predictor of gait economy. When considered across a wide range of walking or running speeds, mechanical power is directly and strongly related \((r > 0.79)\) to walking or running speed (e.g., 45–47) and aerobic demand (45,47). At any given speed, how-
ever, the power-economy relationship is substantially weaker. Taylor and colleagues (38,48) concluded that mechanical work or power does not satisfactorily explain economy variations among species of animals varying widely in body size.

A major limitation of mechanical power computations is their inability to account for isometric contributions of muscles during gait, which recent research suggests may be substantial. In a creative assessment of tendon forces and muscle fiber length changes in running turkeys, Roberts and colleagues (49) reported that most of the muscular force and activity in level running occurs when muscles experience little change in length during the support phase. Operating under near isometric conditions allows a muscle to function in the high force region of its force-velocity curve and thereby tends to maximize force per cross-sectional area of active muscle. Thus, "the demands of support may be met most economically by muscles that produce force while minimizing mechanical work" (p. 1115).

Segment Mass Distribution

The energetic cost of locomotion has been hypothesized to be an important factor influencing the evolution of limb structure in terrestrial animals (e.g., 50–52). This hypothesis is based on the debatable assumption that a major portion of metabolic demand during walking and running is associated with accelerating the limbs with each stride (49). It suggests that those animals that rely on economical locomotion for survival are more likely to reflect a limb structure that presents a smaller inertia load to the musculature, either because less mass is concentrated in the limbs or because limb mass is concentrated more proximally (53). Support for the association between limb inertia characteristics and gait economy comes primarily from segment loading studies in which segment inertia has been modified artificially (e.g., 54–56). The cost of carrying load on the distal aspect of the lower extremities is significantly greater than that when load is carried more proximally on the limb or on the trunk. For example, the relative increase in running aerobic demand produced by adding a given load to the feet is twice as great as that produced by thigh loading (7.0% per kg vs. 3.5% per kg of added load) (54,55), and approximately seven times greater than the increase produced by trunk loading (57). Myers and colleagues (58), however, suggest the mass distribution factor may be of only modest significance. In an assessment of the independent effects of speed, stride rate, and body mass distribution (1.8 kg of mass added to the trunk vs. 0.9 kg of mass added to each shank) on running aerobic demand, they reported that mass distribution accounted for less than 3% of the explained variance in aerobic demand. In contrast, speed and stride rate accounted for 70% and 15% of running aerobic demand variation, respectively. Thus, while the mass distribution hypothesis is theoretically sound, the significance of this factor is not well substantiated or understood.

Joint and Muscular Flexibility

A commonly held belief is that a high level of flexibility is important for optimal running performance. A hypothesis that improved flexibility contributes to less passive resistance about the joints and thereby reduces the demands placed on the musculature is also intuitively appealing. Consistent with this notion, Godes and colleagues (59) reported a reduction in aerobic demand of running following an acute bout of stretching that resulted in increased hip flexion-extension range of motion. In contrast, Gleim and colleagues (60) observed that individuals who displayed poorer flexibility based on a series of 11 static trunk and lower extremity flexibility measures were more economical runners. Their least flexible subjects were as much as 12% more economical than the most flexible individuals. Craib and colleagues (61) reported similar findings. Running aerobic demand was significantly related to maximum passive ankle dorsiflexion angle \(r=0.65\) and standing external hip rotation \(r=0.53\), indicating more economical runners showed more limited ranges of motion for these two joint actions. These two flexibility measures explained 47% of the variance in running economy. Decreased need for activation of posture-stabilizing musculature about the hip and increased elastic energy reutilization about the ankle were speculated to be two mechanisms contributing to more economical running (60,61). Because of the limited empirical research on the association between joint flexibility and gait economy, additional research on the specific effects of flexibility and other structural features on the economy of motion is needed.

GAIT BIOMECHANICS OF SPECIAL POPULATIONS

Gait Adaptations Associated with Aging

There is an extensive body of research literature focusing on changes in gait patterns with advancing age. While the majority of this literature simply describes age-related changes in walking mechanics, there is increasing interest in the potential causes underlying gait changes and in the efficacy of intervention strategies for improving gait mechanics in older adults. The most consistently reported and robust age-related change in gait is a decrease in the preferred or freely chosen speed of walking (e.g., 26,62–67). Preferred walking speeds for elderly adults are typically 10% to 30% lower than those of younger adults. The rate at which walking speed
declines, however, is not constant across adulthood. In an investigation of 438 adults ranging in age from 19 to 102, Himann and colleagues (63) demonstrated that preferred walking speed declines modestly (0.11% to 0.20% per year) until age 60 to 65, at which time the rate of decline of preferred walking speed increases by a factor of 10 (1.24% to 1.60% per year).

Other commonly reported age-related changes in the walking pattern include decreases in stride length and the ratio of swing-to-stance time, and increases in stance width and double support time. An increase in variability of the gait pattern has also been suggested as a possible age-related phenomenon, although this may be a trait linked more specifically with older adults who have a history of falling (68–70). Because many gait parameters are speed dependent and because the majority of research has not contrasted older and younger adults under controlled walking speeds, it is less apparent whether these various temporal and kinematic changes are independent of the decrease in preferred walking speed (64,67). Larish and colleagues (64) reported that older adults demonstrated shorter relative stride lengths than young adults (1.56 vs. 1.66 times leg length, respectively) when both groups were studied at a common intermediate walking speed (1.34 m/s). Thus, their results suggest that age-related changes in gait kinematics are affected by more than just the decline in walking speed.

The underlying causes of the observed slowing of walking with age are less clear. Age-related declines in musculoskeletal function are well documented in the gerontology literature (e.g., 71,72). For example, both dynamic muscle strength and joint range of motion decrease with advancing age (e.g., 73–76). Several studies have reported positive relationships between various expressions of lower extremity muscular strength or power, particularly that pertaining to the ankle plantar flexors, and walking speed in older adults (65,67,77–80). Correlation expressions, however, are typically on the order of 0.4 or lower, indicating that the associations between strength and walking speed are not especially strong. Maximal aerobic capacity has also been positively associated with preferred walking speed (78,81), suggesting that individuals who are more fit aerobically tend to have faster self-selected walking speeds. Similarly, preferred walking speed has also been associated with physical activity status of elderly adults such that those who are more physically active tend to self-select higher speeds (26).

Results of exercise-based intervention programs on gait parameters have been mixed. For example, Cunningham and colleagues (82) reported that 60- to 65-year-old men showed significant increases in both maximal aerobic capacity (9.5%) and preferred walking speed (1.29 to 1.43 m/s) following participation in a 1-year walking or jogging aerobic training program. Using a 20-week multifaceted exercise program (e.g., aerobic, balance, coordination, and strength exercises), Lord and colleagues (83) observed strength improvements (9% to 21%) in several lower extremity muscle groups and a concomitant 6% improvement in preferred walking speed for a community-based cohort of 160 women aged 60 to 83. Similarly, Judge and colleagues (84) reported that 12 weeks of balance and lower extremity resistance training by elderly adults residing in life-care communities produced strength improvements of approximately 30% as well as an 8% increase in preferred walking speed. Other recent investigations, however, have shown no gait improvements in elderly adults following strength and/or endurance training (85–87). Sipila and colleagues (86), for example, reported improvements in both strength and walking velocity in elderly adults following 18 weeks of strength and endurance training. Control subjects, however, reflected similar improvements in walking speed despite the absence of strength gains.

The preceding exercise and aging discussions have focused entirely on physical changes that may be linked to changes in gait. Maki (70) provides an alternative perspective for considering gait changes. He found that slower walking speeds, shorter stride lengths, and a prolonged period of double support were associated with a preexisting fear of falling, but not necessarily with previous history of falling nor subsequent rate of falling. He suggests that age-related changes in gait may actually reflect a strategy to improve stability. This only serves to underscore the importance of examining decrements of gait performance and the efficacy of intervention strategies from multiple perspectives.

Amputee Locomotion

In biomechanical analyses of human locomotion, left-to-right symmetry in all biomechanical variables is typically assumed. The normal variations in segment dimensions (such as girth and strength) appear to have minimal effect on the kinematic and kinetic events associated with either walking or running (88,89). However, for the unilateral below-knee amputee this situation is quite different because the structural asymmetry arising from the amputation results in a concomitant loss of muscle and sensory feedback. The amputee has the substantial challenge of adapting motor control strategies to compensate for the functional loss of significant locomotor musculature and the altered structural properties of the supporting limb. These adaptive processes must incorporate prosthetic limb motion with those of the intact limb to develop a reasonably symmetrical gait (90).

The role of the ankle and foot to functional locomotion is complex. The 52 bones, multiple muscles, and ligaments serve to provide the body with a means to interact with the ground during the support phase. Prior
to 1985 the most common prosthetic foot type was a relatively simple single-axis cushioned heel (SACH) device. More recent prosthetic designs attempt to mimic the motion of an intact foot and lower extremity more effectively, particularly the contributions of the plantar flexors. Winter (91) confirmed the importance of the plantar flexors when he observed that more than 80% of the mechanical power during walking is generated by the plantar flexors during push off. In 1985 the first so-called dynamic energy returning (DER) prosthesis, the Seattle Foot, was released commercially (92). This prosthesis was designed to store energy during early and mid-stance, when the prosthesis was deformed due to normal loading, and to return a portion of that strain energy to the lower extremity during push off. Since the introduction of the Seattle Foot, many other DER prostheses have become available to amputees.

Despite the intuitive appeal of DER designs, research evidence suggests that these prostheses produce only modest effects on biomechanical features of gait. Not surprisingly, these effects appear to be stronger for running than walking because of the higher loading that occurs with running. Gitter and colleagues (93) contrasted walking biomechanics for amputees wearing the Seattle and SACH prostheses. They reported increased energy generation during push-off for the Seattle Foot, but no significant differences in the pattern or magnitude of knee and hip power outputs compared to the SACH foot. Torbourn and colleagues (94) reported no differences in gait symmetry nor in preferred and fast overground walking speed when contrasting several DER prostheses with the SACH foot; they concluded, “There are no advantages of the dynamic elastic response feet for the amputee who is limited to level walking” (p. 383). Czerniecki and colleagues (103) examined the effect of energy-storing prostheses on lower extremity joint moments and muscle power outputs during the stance phase of running. While there were individual differences in response to wearing a so-called dynamic prosthesis, amputees generally showed improvement in power characteristics of the lower extremity when using DER compared to passive prostheses. That is, the dynamic prostheses contributed to power profiles that were more similar to those of nonamputees.

There is little evidence suggesting that DER prostheses lead to substantial metabolic energy savings. Both Torbourn and colleagues (94) and Colborne and colleagues (95) reported no differences in walking aerobic demand when contrasting amputees using DER prostheses and the SACH foot. Postema and colleagues (96) suggested that DER feet would lead to a maximum saving of 3% of metabolic energy during walking. They considered it unlikely that the subjects would notice this difference, and concluded that differences in energy expenditure of this magnitude were probably not clinically relevant.

From a motor control perspective, one of the interesting questions is how unilateral amputees adapt control strategies to offset the substantial losses of structural, motor, and sensory function. Young, active unilateral amputees are capable of generating reasonably symmetrical walking and running patterns from temporal and kinematic perspectives (97-99). Returning to a symmetrical gait pattern following amputation is typically an important goal of the rehabilitation process. Enoka and colleagues (97) observed some differences in running kinematics between their amputee group and previously published data on nonamputees, but they suggested that 60% of the amputee runners exhibited temporal and kinematic patterns that were similar to nonamputees. Furthermore, they showed that the remaining 40% had differences that would have been removed with prosthetic adjustment or training. Sanderson and Martin (98,99) reported marked asymmetries in underlying ankle, knee, and hip moment profiles, but reasonably symmetrical temporal and kinematic features of walking and running gaits. Joint moment asymmetries were particularly notable at the knee and to a lesser extent at the ankle and hip. The contribution of the prosthetic limb knee extensor muscle to the overall support function was substantially reduced or missing, presumably because of an apparent necessity to reduce loading and about the knee. To offset this reduced extensor function, the prosthetic limb hip generally reflected a higher extensor contribution. Miller (100) and Miller and colleagues (101) reported similar adaptations in lower extremity joint kinetics during running.

Not surprisingly, there has been little attention paid to the nonprosthetic leg. Nevertheless, research suggests that intact limb mechanics are altered along with prosthetic limb responses during the motor control adaptation process. Powers and colleagues (102) showed that intact limb loading, as reflected by ground reaction force profiles, was dependent on the type of prosthesis used. Specifically, intact limb loading was reduced with the Flex-foot compared to the SACH foot. Sanderson and Martin (98,99) showed that there were different responses on the intact side compared to the prosthetic side and concluded that there was some modulation of the intact leg to match the prosthetic leg rather than to maintain a pattern similar to the normal. Results suggested that there was no need to develop completely new patterns of joint moments, but rather that it was sufficient to retune the current moment patterns to account for new segmental properties and thus retain internal timing characteristics. Czerniecki and colleagues (103) presented data showing that there was energy transfer between the swing leg and the trunk that may reflect an adaptive strategy to allow energy redistribution to the trunk, which may, in turn, compensate for the reduced power output of the prosthetic limb late in the stance phase.
One of the more recent research emphases relating to prosthetic design and amputee locomotion has focused on the effect of prosthetic limb inertia manipulation on gait biomechanics and energetics. In general, lower extremity prosthetic limbs tend to be much lighter than the limbs they replace, resulting in a substantial inertia asymmetry. This trend toward light prostheses has been driven in part by advances in light-weight materials used in construction of prosthetic limbs and by clinical opinion that prostheses should be as light as possible, presumably to reduce the muscular effort needed for prosthetic control. Data supporting such a prescription, however, are limited. In an early computer modeling effort of the swing leg in which lower-extremity inertia properties were manipulated, Mena and colleagues (104) concluded, “Leg motion was less sensitive to increases in the [segment inertial properties] than to decreases.” They suggested further that “a ‘lightweight’ prosthesis would be less desirable than a ‘heavy’ prosthesis, while a prosthesis that had the same inertial properties of the removed limb may be most desirable” (p. 831). This raise the question of whether inertial modifications of prosthetic limbs may lead to a more effective (e.g., faster preferred walking speeds, more symmetrical walking pattern) and more economical gait pattern.

Czerniecki and colleagues (105) demonstrated that the addition of up to 1.34 kg of mass near the location of the Shank center of mass produced no increases in the aerobic demand of above-knee amputees during walking under multiple controlled speeds. Mean data suggested that aerobic demand tended to decrease slightly with the addition of mass. Similarly, Gailey and colleagues (106) reported no significant change in walking aerobic demand when nearly 1 kg of mass was evenly distributed over the length of the prosthetic limb of below-knee amputees. These results provide encouraging data suggesting that heavier prosthetic limbs do not necessarily lead to higher energy costs. Hale (107) and Gitter and colleagues (108) reported no changes in walking speed and temporal characteristics of the gait cycle as the mass of the prosthetic shank was increased as much as 3.37 kg in above-knee amputees. Both studies showed that increases in shank mass resulted in increases in hip muscular effort during the swing phase, however. Despite this increased demand, four of Hale’s six subjects preferred an intermediate loading condition; that is, one that increased their shank mass, but not to the point such that the prosthetic shank mass matched that of the intact limb. Furthermore, Gitter and colleagues suggested amputees have the ability to self-optimize their gait pattern across a range of prosthetic masses without adverse affects, and that further research needs to be conducted “to determine the psychobiologic underpinnings of changes in perceived exertion associated with alterations in prosthetic mass” (p. 120).

**ELASTIC MECHANISMS DURING WALKING AND RUNNING**

**The Stretch-Shortening Cycle and Musculotendinous Elasticity**

Many human movements, particularly those produced at high speeds and requiring high levels of muscle power, reflect patterns whereby concentric actions of muscle are immediately preceded by eccentric phases. This sequence of active lengthening of muscle followed by active shortening is referred to as the stretch-shortening cycle (SSC) of muscle (109), and it normally results in enhanced muscle performance (e.g., increased muscle work, higher power production) in explosive movements (e.g., maximal vertical jumping). Although this phenomenon is often referred to as a process involving storage and reutilization of elastic energy, it is important to recognize that a variety of mechanisms may contribute to the enhanced performance (see ref. 110 for a review). These include (a) a reflex mechanism in which muscle stretch during the eccentric phase heightens muscle stimulation during the concentric phase (111,112); (b) a heightened level of activation and muscle force present at the initiation of the concentric phase when preceded by eccentric action (113–115); (c) a force-enhancement mechanism whereby stretching of the muscle fibers during the eccentric phase temporarily enhances force-velocity properties (116,117); and (d) an elastic mechanism whereby strain energy is temporarily stored in the musculotendinous unit during the eccentric phase of a movement and then reutilized during the concentric phase (118–121).

Both walking and running reflect SSC behavior in multiple muscle groups, suggesting that muscle output may be enhanced by the mechanisms noted in the preceding paragraph. Neither walking nor submaximal running require maximal production of work or power by active musculature, however. For movements involving repetitive SSCs under submaximal, steady-state conditions, it is more appropriate to consider the benefits of the SSC, particularly the role of elasticity of the musculotendinous unit, in terms of the economy or efficiency of muscle power production rather than work or power generation (110). More specifically, the contractile elements of a muscle do not have to produce all of the necessary increases in mechanical energy during a movement cycle if elastic structures conserve some mechanical energy during an SSC. It should be further noted that only certain musculotendinous units can provide significant elastic energy contributions. Those muscles that have relatively slender and compliant tendons that are substantially longer than the muscle fascicles offer the greatest potential for elastic energy storage and return (122). Ankle plantar flexors are an example. In addition, Alexander and Ker (122) noted that significant metabolic energy savings will occur only if these
relatively compliant tendons undergo large strains (on the order of 4% of their length) under high loading conditions. The need for high stresses to produce these strains is a primary reason that elastic energy contributions in walking are substantially lower than those in running. Hof (123) estimates elastic energy contributions of 13 to 23 J for walking (0.75–1.75 m/s) and 47 to 51 J for running (2.0–2.75 m/s).

There has been some debate on whether the ability of a muscle to store elastic energy can be altered with training and, indeed, whether there is an optimal compliance for effective return of energy. Wilson and colleagues (124) concluded that performance enhancement was increased as a consequence of flexibility training that caused increased compliance, which in turn resulted in an increased utilization of elastic strain energy. The authors further suggested that augmentation of performance from the SSC was dependent on the relationship between the resonant frequency of the movement and the frequency of the SSC itself. In a later study, however, Wilson and colleagues (125) suggested that the rate of force development is a more critical factor for athletic performance than elastic energy contributions, diminishing the value of training-induced changes in elastic response.

Footwear and Surface Elasticity

There is a limited body of research on how elastic properties of running footwear and surfaces influence running performance. The possibility that running shoes have the potential to store strain energy when the viscoelastic materials of the shoe are deformed in early stance and then return some portion of that strain energy to the runner during push-off in late stance has not received significant research support. For a sport shoe to provide a significant return of elastic strain energy to a runner, Nigg and Segesser (126) argued that several conditions must be met. First, the energy must be returned at the location where forces acting at the shoe/foot interface are centered. Second, the energy must be returned at the right time; that is, it must be well synchronized with the push-off at the end of the stance phase. Finally, the energy return must occur at a rate consistent with the rate of development of push-off force by the runner. Using a simple spring model, Nigg and Segesser estimated that under ideal conditions the maximum possible energy return from a sport shoe per step is approximately 5 J or 1% of the total energy per step. Shorten (127) used a much more sophisticated multiple-element, nonlinear viscoelastic model of a running shoe to estimate energy dissipation, storage, and return during the stance phase. Under typical running and footwear conditions, he estimated that 6 to 12 J may be recovered from the shoe. Although he suggested that this amount of energy is potentially significant, he noted that energy storage and return from lower-extremity musculotendinous structures are likely to be more than 10 times greater than that provided by the shoe. Shorten further noted that it may be possible to increase the energy returned to the runner by shifting toward a more resilient sole, which might compromise the cushioning property of the shoe.

McMahon and Green (128) introduced the concept of a “tuned” running surface in which the compliance properties of the surface were tuned to biomechanical characteristics of an individual during the stance phase of running. Their theoretical assessment demonstrated that running surfaces of intermediate compliance can result in a slight enhancement of running speed. Their analysis formed the basis of design characteristics of a new indoor running track at Harvard University. Anecdotal evidence suggested that performance improvements following the installation of the track were modest (approximately 2%), which was consistent with their theoretical prediction. Perhaps equally impressive was the suggestion that running surfaces of intermediate compliance potentially reduce the initial impact peak force, which may contribute to a reduction in injury incidence.

More flexed knee angles at heel strike have also been implicated as an important kinematic mechanism of initial impact shock attenuation. Lafontune and colleagues (129), however, demonstrated that the knee joint does not regulate initial leg stiffness and provided only partial support for their hypothesis that a more flexed knee at impact improves cushioning. Using a human pendulum experimental setup to control impact conditions, they found that greater knee flexion at contact reduced impact force but increased the shock traveling throughout the shank. The impact surface had a larger effect on shock attenuation. More compliant interfaces produced substantial reductions in both initial leg stiffness and severity of the shock experienced by the lower limb. The high correlation ($r = 0.95$) between the rate of lower extremity loading and limb stiffness (defined by the heel fat pad and interface deformations) suggested that interface interventions are more likely to protect the locomotor system against impact loading than neuromuscular strategies involving knee angle at impact.

GAIT ADAPTABILITY

One can consider locomotion to be the translation of the center of mass with the least expenditure of energy possible. Motions of the limbs can be considered to act in harmony to achieve this aim. Saunders and colleagues (130) developed a unifying model that permitted these movements to be classified and in so doing developed the so-called six determinants of gait: pelvic rotation, pelvic tilt, knee flexion, hip flexion, knee and ankle interaction, and lateral pelvic displacement. They ar-
gued that while the body could comfortably account for the loss of one of these determinants, it was much more difficult to do so when two (or more) were lost, due to injury or disease. Loss of two determinants would make effective compensation impossible and the associated metabolic costs would increase to the point that locomotion would not be achievable. We have already seen that with amputee locomotion there is a level of adaptation occurring. Persons who walk with a prosthetic device have shown that they can accommodate many different devices while maintaining overground locomotion (e.g., 94). Furthermore, it seems that an important factor in determining the motion is the desire for symmetrical motion of the lower limbs. Sanderson and Martin (98,99) and Czerniecki and colleagues (103) have shown that the external kinematic features and timing information appear to be retained at the expense of changes within the internal environment. That is, joint moment and powers change but the end result is a temporal or kinematic pattern of motion that is remarkably similar to normal gait.

Adaptations to Surface Constraints

An area of continuing interest concerns the nature of the accommodation, individuals make to changes in the terrain over which they are walking or running. Much effort has been expended examining the importance of vestibular information for balance and postural control (131,132) as well as the visual system for environmental feedback (133–137). Warren (138), Warren and Whang (139), and Patla (140) have focused on identifying specific movement adaptations that are based on features of the environment, such as stair riser heights and obstacles to step over. Patla and colleagues (136) further stressed the importance of the kinesthetic system as one that actively monitors angular displacement of the knee joint and adjusts toe elevation accordingly to ensure obstacle clearance. Specific obstacle clearance strategies were observed, including clearance strategies to go over obstacles of different sizes and shapes safely, implementation of obstacle avoidance strategies within the same step cycle in which the obstacle is presented, and necessary direction changes for avoiding an obstacle, which must be planned in the previous step (135,141). Sanderson and colleagues (142) and Patla and colleagues (143) have observed that when changes to the gait cycle are forced within a step, the adaptation will be made in such a way as to preserve the overall timing of the gait cycle. This appears to be related to the observations made on amputee locomotion, in which adjustments are made to the internal moments and powers to preserve the symmetry of external kinematics.

Gait Plasticity to Biomechanical Training

Based on the information that has been presented for kinematic and kinetic descriptors of gait, one may conclude that the association between biomechanical descriptors of walking and running patterns and gait economy is complex and elusive. Nevertheless, a close examination of published data demonstrates that some individuals clearly do not display economical movement patterns. A particular runner, for example, may tend to significantly over- or understride with each stride. From an energy-conserving standpoint, such individuals may clearly benefit from changes in their pattern of motion. Unfortunately, only a few researchers have attempted to consider whether lasting changes in gait mechanics that significantly affect economy can be produced through biomechanical training.

Results from five recent studies fail to reflect a consensus on the effects of biomechanical training on gait economy and technique. Petray and Krahenbuhl (144) found that the running economy, stride length, and vertical displacement of 10-year-old boys were not significantly improved by an 11-week instruction program that consisted of 5 minutes of instruction per week on various aspects of running technique, including "reducing unnecessary vertical displacement, awareness of stride rate and length, and general suggestions regarding posture and relaxation" (p. 252). Messier and Cirillo (145) reported significant but generally modest changes in gait descriptors, but no significant changes in either oxygen consumption or rating of perceived exertion for female adult novice runners after fifteen 20-minute treadmill training runs during which subjects received visual and verbal feedback on trunk inclination, arm swing, lower extremity mechanics, and vertical oscillation. Miller and colleagues (146) provided visual feedback for a single technique factor to four unexperienced runners during 10 days of 20-minute training sessions. These four subjects displayed reductions in aerobic demand greater than those observed in control subjects, but changes were not statistically significant. Morgan and colleagues (37) provided audio and visual feedback in an attempt to shift nonoptimal stride lengths toward an optimal stride length. Their experimental group showed a significant shift in their freely chosen stride length toward the optimal stride length relative to a control group without feedback. Furthermore, they also showed a marked reduction in the oxygen uptake from that recorded at the freely chosen stride length. Finally, Williams and colleagues (147) evaluated the effect of training at a longer than optimal stride length on numerous running technique descriptors and aerobic demand. Posttraining technique descriptors were generally intermediate to those of the pretraining and imposed training values, but most kinematic adaptations were not significant. Posttraining aerobic demand was significantly less than pretraining values, a change opposite to that which would be expected. The O2 decline also was not specific to the training stride length but rather occurred across all tested stride length conditions.
From these analyses, it appears that both economy and gait technique factors are resistant to change due to training programs such as those described. Perhaps these mixed results should not be surprising considering the brief duration of the training programs, differences in experimental design, the highly complicated nature of the interrelationships among various technique factors, and our relatively poor understanding of factors believed to be primary determinants of interindividual differences in economy of motion. The question regarding our ability to improve economy significantly through biomechanical training remains unanswered and warrants further research.

**Gait Transitions**

One of the most basic locomotor adaptations made by humans and animals is the transition from one mode of gait to another (e.g., walk to run, trot to gallop) in response to changes in speed. Numerous investigations have attempted to identify critical factors that trigger transitions from one mode to another. Proposed mechanisms have focused on minimization of energy cost, minimization of musculoskeletal stress, limitations in the stride pattern imposed by anatomic or biomechanical constraints, minimization of mechanical energy or power generation, and changes in the stability of the stride pattern.

Hoyt and Taylor (148) demonstrated that the energy cost per unit of distance traveled for horses reflects a U-shaped, curvilinear response within each mode of gait (walk, trot, gallop) such that the highest costs were observed at the extreme speeds for a given mode of gait. They also observed that preferred speeds within each gait mode coincided with the energetically optimal speed. Although they did not systematically quantify gait transition speeds in their analysis, they concluded that changes in gait occur in an effort to minimize energy demand. In tests on human subjects, both Hreljac (149) and Brisswalter and Mottet (150) questioned the importance of energy cost minimization when they found that the predicted transition speed based on minimization of energy cost (2.24 and 2.19 m/s for Hreljac and Brisswalter and Mottet, respectively) was significantly greater than the observed walk-to-run transition speed (2.11 and 2.13 m/s, respectively). Whether this difference between the preferred and the energetically optimal transition speeds is of physiological significance remains open to interpretation and debate.

Farley and Taylor (151) discounted the role of energy cost in triggering gait transitions and proposed that musculoskeletal forces were a more reasonable triggering mechanism in part because of the availability of biologic transducers for detecting and communicating loading information to the central nervous system. Focusing on the trot-to-gallop transition in horses and using vertical ground reaction force data to predict forces of distal extensor muscles, they concluded that the gait transition occurred when musculoskeletal forces reached a critical level. This same mechanism, however, does not explain the walk-to-trot transition in quadrupeds nor the walk-to-run transition in humans because contact forces, and thus musculoskeletal loading, tend to increase with these transitions (152,153).

Several investigators have considered whether anatomic (e.g., leg length) or biomechanical (e.g., stride length) constraints affect gait transition (e.g., 154–156). Minetti and colleagues (156) concluded that the angle of lower limb spread (i.e., maximum inter-thigh angle) is an important anatomic constraint that limits walking step length and thereby stimulates a walk-to-run transition. This conclusion, however, was not generalizable to the run-to-walk transition during which each high-angle leg increased, nor to walk-to-run transitions under different incline conditions where maximum inter-thigh angle changed as incline changed. Hreljac (152) proposed that fatigue or exertion levels in the ankle dorsiflexors trigger the walk-to-run transition. Under different incline conditions, he observed that the walk-to-run transition speed increased as incline increased. The maximum ankle dorsiflexion velocity, however, reached the same peak value under the different incline and transition speed conditions. Hreljac proposed that perceptions of localized discomfort in the dorsiflexors is a plausible source of feedback to the central nervous system that prompts a gait transition. Unfortunately, the run-to-walk transition cannot be explained by this same mechanism.

Using a simple inverted pendulum model of walking (157), Kram and colleagues (153) suggested that the walk-to-run transition may occur in an effort to minimize the demand for mechanical power generation. At preferred walking speeds, substantial exchanges between kinetic and gravitational potential energy components from a pendulum-like mechanism minimize the need for power generation by musculature to sustain walking. As walking speed increases, however, the effectiveness of this mechanical energy exchange diminishes. Under running conditions, minimization of mechanical power generation is associated with an elastic spring mechanism, rather than a pendulum energy exchange mechanism. Thus, it can be suggested that the walk-to-run transition occurs at a speed at which the energy-conserving characteristic of the elastic mechanism exceeds that of a pendulum mechanism. Theoretically, the only major force affecting the inverted pendulum-like movements during walking is gravity. By supporting the body vertically at varying levels and thereby simulating reduced gravity conditions, Kram and colleagues found support for their hypothesis that the walk-to-run transition occurs at slower absolute speeds as the gravitational effect is reduced. The effects of reduced gravity on the
relative effectiveness of pendulum-like and elastic spring mechanisms at conserving mechanical energy was not tested in these experiments, and is unclear.

A final potential mechanism underlying gait transitions deals with the stability of the gait pattern. Dynamical systems theory (158) predicts that complex systems, such as the locomotor process in humans and animals consisting of a high number of degrees of freedom, are self-organized into a small number of simple and distinctive patterns (e.g., walking and running). Based on dynamical systems theory, it would be predicted that preferred speeds of walking and running reflect stable attractor states of the motor system. Increases in walking speed or decreases in running speed from the preferred states should result in less stable and more variable kinematic and kinetic attractor states of gait that trigger a transition. The majority of research to date on dynamical systems theory has focused on simple upper extremity movements. From a gait perspective, researchers are still attempting to identify those collective variables that most effectively encompass multiple degrees of freedom. Nevertheless, using relative phase differences between hip, knee, and ankle peak extensions near the end of the stance phase, Dietrich and Warren (159) effectively manipulated attractor state stability through inertial and incline or grade manipulations and correspondingly altered the walk-run transition speed. In addition, Brisswalter and Mottet (150) observed an increase in stride duration variability as walking speed approached gait transition speed. With a further increase in speed and a shift to running, stride duration variability abruptly dropped.

The preceding discussion of gait transition reflects many perspectives from which the phenomenon has been investigated, and it perhaps underscores the complexity of the locomotor process. It may also leave the reader with the mistaken impression that the various perspectives are mutually exclusive. Rather, it is more likely that gait transition is a multifactorial issue and that some of the energetic, biomechanical, and anatomic factors discussed as important determinants of gait transition are interrelated. As one example, Dietrich and Warren (159) speculated that gait stability and metabolic energy cost are closely linked, suggesting that the "metabolic cost reflects the consequences of driving the system away from its attractor states" (p. 61).

**CONCLUSION**

This chapter provided an overview of our current state of knowledge on a limited number of topics, in part because a broader review would lead to a more superficial consideration of topics and because many other reviews already exist on various topics associated with walking and running biomechanics. It is hoped that our discussion has reinforced the notion that walking and running are extremely complex movements that are best understood only when investigated thoroughly from multiple perspectives that include, but are not limited to, kinematic, kinetic, metabolic, neurobiologic, and psychobiologic assessments.

**REFERENCES**


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