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Review

Biomechanics and muscle coordination of human walking Part II: Lessons from dynamical simulations and clinical implications

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Abstract

Principles of muscle coordination in gait have been based largely on analyses of body motion, ground reaction force and EMG measurements. However, data from dynamical simulations provide a cause-effect framework for analyzing these measurements; for example, Part I (Gait Posture, in press) of this two-part review described how force generation in a muscle affects the acceleration and energy flow among the segments. This Part II reviews the mechanical and coordination concepts arising from analyses of simulations of walking. Simple models have elucidated the basic multisegmented ballistic and passive mechanics of walking. Dynamical models driven by net joint moments have provided clues about coordination in healthy and pathological gait. Simulations driven by muscle excitations have highlighted the partial stability afforded by muscles with their viscoelastic-like properties and the predictability of walking performance when minimization of metabolic energy per unit distance is assumed. When combined with neural control models for exciting motoneuronal pools, simulations have shown how the integrative properties of the neuro-musculo-skeletal systems maintain a stable gait. Other analyses of walking simulations have revealed how individual muscles contribute to trunk support and progression. Finally, we discuss how biomechanical models and simulations may enhance our understanding of the mechanics and muscle function of walking in individuals with gait impairments.

Keywords: Locomotion; Musculo-skeletal system; Computer models; Motor control; Neural control

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1. Introduction

Observation of adult human walking suggests that coordination of this task is seemingly simple. After all, walking 'appears' quite graceful. Nevertheless, how individual muscles fulfill the mechanical requirements of the locomotor task remains controversial even though studies of gait have a long history [2,3].

Analyses of measurements obtained from individuals while walking are now quite common. Muscle electromyographic (EMG) activity can be measured with surface or indwelling electrodes [4–7] along with measurements of the kinematics of the body and the ground reaction forces [8]. Laboratories devoted to gait studies are common and typically well instrumented to record these data [9]. The position, velocity and acceleration of markers placed on the body can be measured or estimated while subjects walk, and the ground reaction force measured with force-plates or just the vertical component with pressure sensors in shoe soles [9,10]. In addition, other biomechanical variables can be estimated from the body segment kinematics and the ground reaction force measurements using inverse dynamics, such as body segmental energetics and the net joint moment and power at the ankle, knee, and hip [11–13]. The accuracy and reliability of the techniques utilized in these measurements and calculations have been assessed [14–22]. Over the years, such measures of biomechanics and muscle activity of walking have led to the publication of kinematic, kinetic, and muscle activity trajectories over the gait cycle for 'healthy' young and older adults [23,24] and children [25], though more data for subpopulations of the elderly and patients with movement abnormalities are still needed [24]. Compila-

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tion of standardized gait data across laboratories may be the solution to acquiring such databases [26–28].

Clinical gait laboratories often compare the walking characteristics of individuals with orthopaedic or neurologic pathologies to the walking patterns of healthy individuals [25,29,30]. This comparison of the walking kinematics, kinetics and muscle activity patterns between healthy and impaired individuals serves as the basis for defining abnormal gait [31–33]. Such comparisons can also be used for neuromuscular and musculoskeletal gait assessment, surgical and rehabilitation treatment, prosthetic design and fitting, and even to the development of Functional Electrical Stimulation (FES) systems for individuals with spinal cord injury [34–36]. The comparison of walking in the healthy elderly to healthy young adults, for example, has revealed that hip flexion contractures and ankle plantar flexor weakness can limit performance, independent of gait speed [37]. On the other hand, decreased walking speed was found to be an important variable in describing hemiparetic gait when post-stroke individuals were compared with a healthy control group [38]. Regression analyses have been used to identify associations between temporal, kinematic and kinetic variables with performance criteria, such as walking speed [39]. The effect of exercise on the degree of chronic impairment and disability in individuals with post-stroke hemiparesis has been evaluated using various gait outcome measures [40].

Critical to the success of using gait laboratory measurements in the medical treatment of individuals with neuro-musculo-skeletal pathology is the skill of the surgical, clinical and engineering gait-laboratory team to interpret the measurements from a specific patient [27,41]. At the moment, the team's ability is based on its 'hands-on' experience. The cause of the deviation of a patient's walking pattern from the healthy pattern is inferred from the correlation the team has made in the past between walking pattern deviations and its treatment protocols and decisions. However, treatment protocols often differ among laboratories.

We posit that the causal relationship between EMG patterns and gait kinematics and kinetics has to be ascertained before there can be a clear understanding of how muscles coordinate walking in individuals. Further, it is believed that the delineation of such relationships between neuro-musculo-skeletal pathology and abnormal walking patterns will lead to more successful and perhaps new surgical and rehabilitation treatments. Then the gap between the theoretical framework used in common medical practice and that used for gait analysis in clinical laboratories [42–44] can be narrowed [36,45].

Dynamical simulations driven by individual muscles are believed to be critical to understanding the causal relationships between EMG patterns and gait kinematics and kinetics (see Part I [1]). The foundation for generating simulations will no doubt rely on interactive computer-implemented musculo-skeletal models [46–49] because of our inability to measure the necessary biomechanical quantities directly. It is more challenging to generate simulations of walking than other tasks, such as pedaling (see Part I [1]), because in walking the pelvis is neither stationary nor externally supported making the system highly unstable, and the interaction between the feet and the ground is complex [50–52]).

Part II reviews how simple and complex dynamical models and simulations have provided insight into the biomechanics and coordination of walking, including the passive and ballistic properties that emanate from the inverted multisegmented body, the ability of the force–length–velocity property of muscles to stabilize gait, the role of the uni- and biarticular ankle plantar flexors to support the body and provide forward progression, and the importance of muscle force generation to the redistribution of segmental energy irrespective of whether the muscle is shortening, lengthening, or neither. Muscle function and coordination of forward progression and vertical support in the sagittal plane is emphasized (cf. hip abductors to maintain frontal plane support). The reader is referred to other reviews of walking that discuss the measurements of kinematics and ground reaction forces, and the net joint moments, net joint powers, and segmental energetics derived from inverse dynamics [10,23,32,35,53–55].

2. Simple mechanical models: basic energetic principles of bipedal walking

2.1. Determinants of gait, inverted pendulum and mass-spring models

A first approach to understanding a complex locomotor task, such as human walking, is to understand the basic mechanics of the multi-segmented body using simple mechanical models without muscles. Clearly, the less complex a walking model, the easier it is to analyze and gain insight into fundamental mechanisms [56]. For example, six determinants of gait have been defined from a set of simple physical models and used to qualitatively describe how the motion of the pelvis, legs and feet provide for a smooth motion of the body center-of-mass [35,57]. These models are seemingly insightful and often referred to summarily in reviews of gait [58]. Nevertheless, the qualitative use of such simple kinematic models to account for the complex dynamics of walking will lead to erroneous conclusions at times [59]. Gard and Childress [60,61], noting that the presumed relationships between the determinants and gait kinematics have not been scrutinized experimentally, have recently shown that the second (pelvic list, or lowering of the pelvis on the swing-leg side) and the third determinant (stance-leg knee flexion) have little effect on trunk excursion, in conflict with the original descriptions of the determinants-of-gait. Joint-moment driven dynamic simulations using multi-linked 3D leg models also suggest that pelvic list is less important than conventionally assumed [62].

The inverted pendulum model is an excellent example of a simple model providing insight into the basic energetics of walking. By assuming the leg is very stiff and extended during mid stance in walking, the mechanics of the inverted pendulum model is compatible with the transfer of body kinetic energy into potential energy then (for review, see Ref. [63]). However, the inverted pendulum model cannot account for the two-peak vertical ground reaction force observed in human gait. Inclusion of stance-leg flexion and foot–knee interaction in a model can [64,65], though other model modifications may also. Our understanding of basic running energetics has also been considerably enhanced by simple models; in this case by mass-spring models, where body mass is assumed to reside at the pelvis and the legs assumed to act like springs. Inverted pendulum models of walking and mass-spring models of running are powerful in understanding the transition from one gait to another, for example, from walking to running [63,66–68]. However, inverted pendulum and mass-spring models do not attempt to account for the multi-linked nature of the legs and provide little insight into muscle coordination principles.

2.2. *Ballistic and passive walking models*

Limited degrees-of-freedom dynamic walking models are especially powerful in understanding how dynamic interactions among the legs, and the segments within a leg, can exploit the inherent dynamical properties of the multi-linked system to generate efficient locomotion. Mochon and McMahon [69,70] showed that body progression during swing, and thus single-leg support, can be accomplished in the absence of any muscle force generation for a reasonable set of kinematic conditions at toe-off because of the potential and kinetic energy exchanges between the swing-leg and the trunk, and between the trunk and the stance-leg (however, see Ref. [71]). Thus, they referred to this type of walking as ‘ballistic walking’.

Other studies with conceptually-similar planar anthropomorphic walking models, with or without knees, have shown that periodic stable walking can be achieved without any muscular action whatsoever when descending down a gentle incline because gravity compensates for the kinetic energy lost during ‘heel strike’, and during knee hyperextension near the end of swing if the model has knees [72–74]. Thus, the dynamics of this simple walking model allows for a periodic walking cycle down a slight incline to be established without any external energy input except for gravity. This type of walking is referred to as ‘passive walking’.

Using the simplest passive walking model [75] and other more anthropomorphic passive walking models with active torque actuators to allow walking on level ground [74], Kuo [76] showed that an actuator-supplied impulse of energy immediately before toe-off, for example, to mimic plantarflexor-produced energy before toe-off, is particularly effective because energy loss at ‘heel strike’ of the contralateral leg is minimized. Similarly, van der Linde [77] showed that stable ballistic walking on flat ground can be achieved with simple phasic muscle contractions controlling hip torque and leg extension. Using a passive walking model to study the speed–step length relation under different metabolic cost assumptions, Kuo [78] showed that a metabolic cost associated with both push-off near toe-off, and the amount of force needed to oscillate the swing-leg per unit time applied, for example to mimic hip flexor activity in early swing, predicted well the speed–step length relationship observed in humans.

Extending the planar passive walking models to 3D so that frontal plane motion is allowed, Kuo [79] also showed that passive walking cycles exist, as in the planar walking models, but medial–lateral motion is unstable, though stability can be attained with quasi-static control of step width. The dynamic model and control predictions regarding variability in fore-aft and lateral foot placement and the coupling of step width to length were confirmed experimentally in human subjects, leading Bauby and Kuo [80] to conclude that humans may harness much of the passive, inherently stable dynamics of the legs in the sagittal plane, but probably provide significantly more active muscle control to stabilize motion laterally.

2.3. *Limitation of simple mechanical models*

Though these simple, non-muscle based models are extremely amenable to analysis and provide much insight into the basic relationships among body height and mass, leg stiffness and length, walking speed, and overall mechanical energetics, a significant limitation in all of them is their inapplicability to understand muscle coordination of walking. The reason is that these models do not have anthropomorphic muscles, nor can they, virtually by definition. The inability of these models to understand muscle coordination in double-leg support is especially noteworthy because many muscles are active then [4,23,32].

Only a few attempts have been made to add ‘muscles’ to these simple models [68,77]. Assuming a simple mechanical model with a knee and two hypothetical actuators with muscle-like properties, Minetti and Alexander [68] showed that human stride length and the speed of transitioning into a run are predictable. Their simulations also reproduced the ground reaction forces, even during double-support. But the simulations were, in effect, constrained to do so because two parameters in the model, which are the shape of the ground reaction force and the swing/stance time ratio, were adjusted so the simulation data would match the experimental data and have a realistic double-support phase. The fulfillment of these constraints requires the coordination of additional muscles, which in turn might affect the putative role of the two muscle-like actuators in the model. A better understanding of muscle coordination requires, therefore, a model with enough muscles to represent all the major lower extremity uni- and biarticular muscle groups.

3. Dynamical models and simulations of gait actuated by net joint moments

Simulations from dynamical multi-segment models driven by net joint moments provide some insight into the function of muscles. With simulations driven by net joint moments, the instantaneous contributions of each net joint moment to the acceleration of the body segments, the joint intersegmental forces, and the energy flow in/out of the segments can be found (see Part I [1]). Because net joint moments are produced largely by muscles, dynamical models actuated by net joint moments found from traditional Newton–Euler inverse dynamics provide some insight into the causal relationships between muscle forces and segmental kinetics.

Kepple et al. [81] studied the instantaneous contribution of the net ankle, knee, and hip moments to the acceleration of the trunk in the forward and upward directions, considered analogous to forward progression and vertical support, respectively. They used a dynamical model actuated bilaterally by net joint moments in the leg, where the moments were derived from traditional inverse dynamics. The body model had seven segments, which were two feet, two shanks, two thighs, and a trunk. Kepple et al. analyzed the second half of single-leg stance and concluded that the ankle plantar flexors are the primary contributors to forward progression and vertical support. Muscle-based simulations of walking support this contention, though other muscles [82], probably the uniarticular knee and hip extensors [83], may provide substantial forward progression and support in the beginning of stance, a region not studied by Kepple et al.

It has long been recognized that non-muscle based dynamical models might be useful to the understanding of mechanisms contributing to pathological gait. Both 2D and 3D dynamical models of part or all of the gait cycle driven by net joint moments have been used to understand pathological gait [62,84] and assistance provided by orthoses [85], to plan therapy and surgery [86], and to design feedback control for FES systems [87]. These earlier models seem to have been used little later, probably because the computational time in producing simulations had been very high, and are still constraining (personal experience).

The recognition of the usefulness of non-muscle based dynamical models to understand pathological gait continues [88]. Using a two leg, trunk model driven by inverse-dynamics computed ankle, knee and hip moments, Kerrigan et al. [89] showed the feasibility of using dynamical models to understand patient-specific mechanisms of adult spastic paretic stiff-legged gait in swing. In studying the whole gait cycle of stiff-legged-gait patients, Riley and Kerrigan [90] concluded similarly that dynamical models and induced acceleration analyses have much potential in understanding the link between the neurological impairment and the specific walking disorder in a patient.

Caution is advised in interpretations of function of *individual muscles* using dynamical models actuated by net joint moments. Because biarticular muscles are active simultaneously with uniarticular muscles, the function of an individual muscle cannot necessarily be deduced from the effects found from analysis of the action produced by the net joint moments.

4. Dynamical simulations: effect of muscles on gait production and stability

Simulations derived from dynamical models with physiologic-like muscles have very high potential for understanding muscle coordination of gait [63,91–93]. However, the challenge is great [27].

Yamaguchi and Zajac [94] have suggested that only seven major muscle groups have to be stimulated to restore near-normal walking in persons with paraplegia based on simulations generated from a 3D gait model of single-leg stance. They heuristically analyzed the resulting nominal simulation of normal gait by observing the development of pathological characteristics in the simulation when changes were made to the nominal muscle excitation pattern. Perhaps their most important conclusion was that control of gait must be synthesized, in the case of FES systems, or analyzed, in the case of understanding gait characteristics, on a system wide basis because the force in each muscle affects the acceleration of each body segment. Other dynamical models, with only uniarticular hip and knee muscles however [95], have been developed to design learning algorithms for FES control of walking [96].

Anderson and Pandy [97] have concluded that minimization of metabolic energy expenditure per unit distance traveled is a valid predictor of walking performance. Rather than apply dynamic optimization to their dynamical model to replicate experimental gait kinematics and kinetics, they used dynamic optimization to predict the muscle excitations, and thus the resulting gait trajectory, subject to minimization of an assumed task objective. Their 3D model has 23 degrees-of-freedom and is actuated by 54 muscles [98]. Much processor time of a supercomputer was required to solve the optimization problem. The solution produced reasonable gait kinematics and ground reaction forces.

Gerritsen [99] and Gerritsen et al. [100] showed that the intrinsic force–length–velocity properties of muscles facilitate stability during walking. They generated four nominal simulations using a seven-segment model of the legs and the trunk with 16 muscles per leg providing the actuation. Though each nominal simulation emulated the same natural walking movement, four different models for the muscles were used. The different muscle models had the following properties: only force–length, only force–velocity, neither, or both. The four nominal simulations were subjected to static and dynamic perturbations, such as a heavier trunk, increased gravity, or a vertical or horizontal impulsive force applied to the trunk in double-support. The walking movement was least disturbed, at times hardly at all, when muscles had both a force–length and a force–velocity property.

5. Neural control of walking using neuro-musculo-skeletal models

Dynamical models of the musculo-skeletal system combined with neural control models for the activation of muscles have been developed to simulate the integrative properties of the neuro-musculo-skeletal system in the maintenance of stable gait. Neuro-musculo-skeletal models are useful in developing computational theories of motor control [101,102]. A noteworthy neuro-musculo-skeletal model has been generated by Taga and coworkers [102–104]. Their musculo-skeletal model has eight body segments, viscoelastic ground contact elements, and 20 muscles but without activation or contractile dynamics. Their neural control model is composed of an impedance controller, a sequence of global states that depend on the phase of the gait cycle, a rhythm generator exciting seven neural oscillator pairs, with one for the trunk and two each for the hip, knee, and ankle joints, and sensory feedback providing input to the impedance controller, the global state controller and the generator.

Taga et al. [102–104] showed that gait emerged as stable limit cycle through global entrainment of the neural system and the musculo-skeletal system including its interaction with the ground. Further studies showed that walking movements are reasonably robust against mechanical perturbations, trunk loads, and variations in the terrain, that speed can be controlled by a single parameter tonically exciting the neural oscillators, and that step cycle can be entrained by a rhythmic input to the oscillators [105]. Presumably this model would have been even more stable against perturbations had it included the force–length–velocity property of muscle (cf. [100]). Obstacle avoidance, which is the maintenance of gait when changes in the environment occur rather fast compared to the walking rhythm, was possible with the addition of another neural processing component that receives visual input and interacts with the rhythm generator and the motor output [106].

Another approach to the synthesis of human gait using neuro-musculo-skeletal models is to emphasize the control afforded by a central pattern generator, the dynamic properties of muscles, and reflexes from muscle spindles, tendon organs, and foot tactile receptors [107,108]. Using a genetic algorithm to find the neural connectivity weights in a model that minimized energy consumption per step, Ogihara and Yamazaki [107] found multiple step walking to be possible, though less robust against perturbation than the Taga et al. model. Nevertheless, using this model to study the ontogeny of human walking, Yamazaki et al. [109] suggest that development of upright walking is characterized by a change in reliance on force oscillation of the body through neural control to reliance on pendular motion of the body. Interestingly, the tendon organs were found to participate little to the motor output, perhaps because they were assumed to inhibit rather than enhance homonymous motor activity, as occurs during locomotion [110,111].

None of these studies analyzed the dynamic simulations to understand how individual muscle forces affect the acceleration of the segments or contribute to the segmental energetics. Since simulations were generated, analyses could have been performed to understand the biomechanics. Instead, their studies emphasized neural control.

6. Understanding muscle coordination of walking with dynamical simulations

Building on the pioneering work of Yamaguchi [112], recent analyses of simulations of the whole gait cycle have begun to identify the instantaneous contributions of individual muscles to the acceleration and energetics of the individual body segments [82,113]. Analysis of these muscle-induced segmental accelerations and powers have shown how muscles work together in synergy, or independently to coordinate movement of the body [82].

6.1. Determining muscle contributions to segmental acceleration and power

Recent analyses of gait simulations to determine the ‘instantaneous’ contributions of individual muscles to segmental accelerations and powers have attempted to include the additional effects produced by individual muscles on the ground reaction force, specifically when foot/ground contact is modeled by a set of viscoelastic elements [82,113]. Without loss of generality, say the dynamical equations of motion are given by:

$$\ddot{\boldsymbol{q}} = \mathbf{I}^{-1}(\boldsymbol{q}) \times \{\mathbf{R}(\boldsymbol{q})\boldsymbol{F}^{\text{mus}} + \boldsymbol{F}^{\text{grf}}(\boldsymbol{q}, \dot{\boldsymbol{q}}) + \boldsymbol{G}(\boldsymbol{q})\boldsymbol{g} + \boldsymbol{V}(\boldsymbol{q}, \dot{\boldsymbol{q}}) + \boldsymbol{F}^{\text{non}}(\boldsymbol{q}, \dot{\boldsymbol{q}})\}, \tag{1}$$

where $\boldsymbol{q}$, $\dot{\boldsymbol{q}}$, $\ddot{\boldsymbol{q}}$, vector of generalized coordinates, velocities and accelerations (e.g. segment angles, angular velocities, angular accelerations); $\mathbf{I}(\boldsymbol{q})$, system mass matrix; $\mathbf{R}(\boldsymbol{q})\boldsymbol{F}^{\text{mus}}$, vector from net muscle moments ($\mathbf{R}(\boldsymbol{q})$, moment arm matrix, and $\boldsymbol{F}^{\text{mus}}$, vector of muscle forces); $\boldsymbol{F}^{\text{grf}}(\boldsymbol{q}, \dot{\boldsymbol{q}})$, vector from ground reaction forces; $\boldsymbol{G}(\boldsymbol{q})\boldsymbol{g}$, $\boldsymbol{V}(\boldsymbol{q}, \dot{\boldsymbol{q}})$, $\boldsymbol{F}^{\text{non}}(\boldsymbol{q}, \dot{\boldsymbol{q}})$, vector from gravity, Coriolis and centripetal, and non-muscle forces (other than from the ground reaction force).

Then the contribution of an individual muscle force F_i^{mus} to the ‘instantaneous’ acceleration of the segments $\ddot{\boldsymbol{q}}$ at some instant is presumed to be the summed contribution arising from F_i^{mus} at that instant, which appears in the first term on the right side of Eq. (1), and the ground reaction force due to the immediate past trajectory of F_i^{mus}, which is a component of the second term in Eq. (1). The rationale for including the muscle’s effect on the ground reaction force is that the ground is considered to be rather ‘stiff’ compared to the segmental dynamics and to $\boldsymbol{F}^{\text{non}}(\boldsymbol{q}, \dot{\boldsymbol{q}})$; thus an applied force over a short interval, like from a muscle or from gravity, may change the ground reaction force much even though the change in the positions and velocities of the segments will be small.

The challenge is to determine the preceding time interval and methodology appropriate to the estimation of the effects of the past trajectory of F_i^{mus} on the ground reaction force at each instant. Neptune et al. [82] estimated the contribution of a muscle force to the force in each foot contact element by removing one muscle force from the simulation over a 2.2 ms time interval and computing the change in force in each foot contact element. The change in force was then presumed to correspond to the contribution of that muscle to the foot contact force. Sensitivity analyses were performed to ensure that interpretation of individual muscle function was insensitive to the time interval used in the calculation. They also ensured that the power delivered to all the segments by a muscle equals the power produced by the muscle. On the other hand, they did not include the contribution of muscle power to the foot contact elements, which perhaps should be considered since these elements store and dissipate energy also.

Other investigators have neglected to consider the contribution of a muscle force to the ground reaction force [90] or have assumed that the foot-ground contact could be simplified and represented by a joint [81]. When the foot-ground contact is represented as a joint, the muscle contribution to the ground reaction force is determined like any other joint reaction force in the model. Recognizing the importance of decomposing the ground reaction force, Anderson [113] has also considered the contributions to the ground reaction force by using hard kinematic constraints to decompose the ground reaction force.

It has become clear that the problem of decomposing the ground reaction force into individual muscle contributions is a major obstacle to the determination of the ‘instantaneous’ contributions of muscle forces to segmental acceleration and power. Perhaps, one should just recognize that the instantaneous contributions necessary for an induced segmental acceleration and power analysis are those arising from only $\mathbf{R}(\boldsymbol{q})\boldsymbol{F}^{\text{mus}}$. The challenge would then be to find methods to account for the total contribution of the past trajectory of, say, a muscle force to the movement of each body segment, and thus its contribution to $\boldsymbol{F}^{\text{grf}}(\boldsymbol{q}, \dot{\boldsymbol{q}})$, $\boldsymbol{V}(\boldsymbol{q}, \dot{\boldsymbol{q}})$, $\boldsymbol{F}^{\text{non}}(\boldsymbol{q}, \dot{\boldsymbol{q}})$ at each instant as well (see Part I [1]).

6.2. Generating a dynamical simulation

In Part I [1], we reviewed the two conceptual methods for generating dynamical simulations. One method is to define a priori the task objective. Anderson and Pandy [97] assumed the objective in walking to be minimization of metabolic energy expenditure per unit distance traveled and found overall walking performance to be predicted well when muscle excitations were adjusted to achieve this objective. The other method is to solve the optimal tracking problem by adjusting the muscle excitations to replicate the experimental walking data as well as possible [82] regardless of whether energy is minimized to the extreme or not. We believe the latter approach is most promising to the study of a variety of gait patterns, including pathological gait and subject-specific gait.

Because Neptune et al. [82] used this latter approach to study walking of young healthy adults, and their results are referred to frequently below, their methodology is now described in some detail. They generated a dynamical simulation that replicated the kinematics and kinetics of walking at 1.5 m/s using a sagittal-plane musculo-skeletal model having two legs and one ‘trunk’ segment, which represented the head, arms, torso and pelvis. Each leg had a thigh, shank, patella and foot, and was actuated by 14 muscles composed of Hill-type properties and in-series elastic structures, such as a tendon [114]. Foot/ground contact was modeled by 30 stiff, lightly damped viscoelastic elements [115]. An algorithm [116] determined the on-off timing and magnitude of the muscle excitations that minimized the difference between simulated kinetic and kinematic sagittal-plane trajectories and gait lab measurements. The kinetic/kinematic trajectories were the right and left hip, knee and ankle joint angles, the net joint moments and powers, the horizontal and vertical ground reaction forces, and the horizontal and vertical translation of the trunk. Each simulated trajectory was near ±2 SD of the corresponding experimental kinetic/kinematic trajectory (Fig. 1), including the anterior/posterior ground reaction force, which contributes significantly to total metabolic cost, at least in running [117]. Simulated phasing of muscle excitations compared well with published EMG information (Fig. 1) and body kinematics were realistic (Fig. 2). Other simulations were generated with muscle excitations varied somewhat. The model was sufficiently complex to elucidate the basic muscle coordination mechanisms of forward progression and sagittal-plane vertical support, yet simple enough to be computationally tractable.

6.3. *Contribution of the uni- and biarticular plantar flexors to support and forward progression*

Neptune et al. [82] found that the uni- (Soleus, SOL) and the biarticular (Gastrocnemius, GAS) plantar flexors begin to support the trunk in early single-leg stance as the support provided by other muscles wanes, and the plantar flexors alone support the trunk during the rest of single-leg stance into pre-swing (Fig. 3 *Trunk Support*). The trunk is supported by the plantar flexors because their individual contributions to the hip intersegmental force have a large vertical component, which accelerate the trunk upwards before mid stance ($<30\%$ of the gait cycle) and afterwards decelerate the downward motion of the trunk (Fig. 4). Thus, the individual plantar flexors increase the vertical energy of the trunk before mid stance and decrease vertical trunk energy afterwards.

In contrast, SOL and GAS were found to contribute to trunk forward progression only from mid stance through pre-swing; before mid stance, they hinder progression (Fig. 3 *Trunk Forward Progression*) [82]. But the contribution of SOL is larger (Fig. 4, 30 *and* 45% *Cycle:* compare forward components of hip intersegmental force induced by SOL and GAS). Thus SOL increases the horizontal energy of the trunk much more than GAS, especially in late stance.

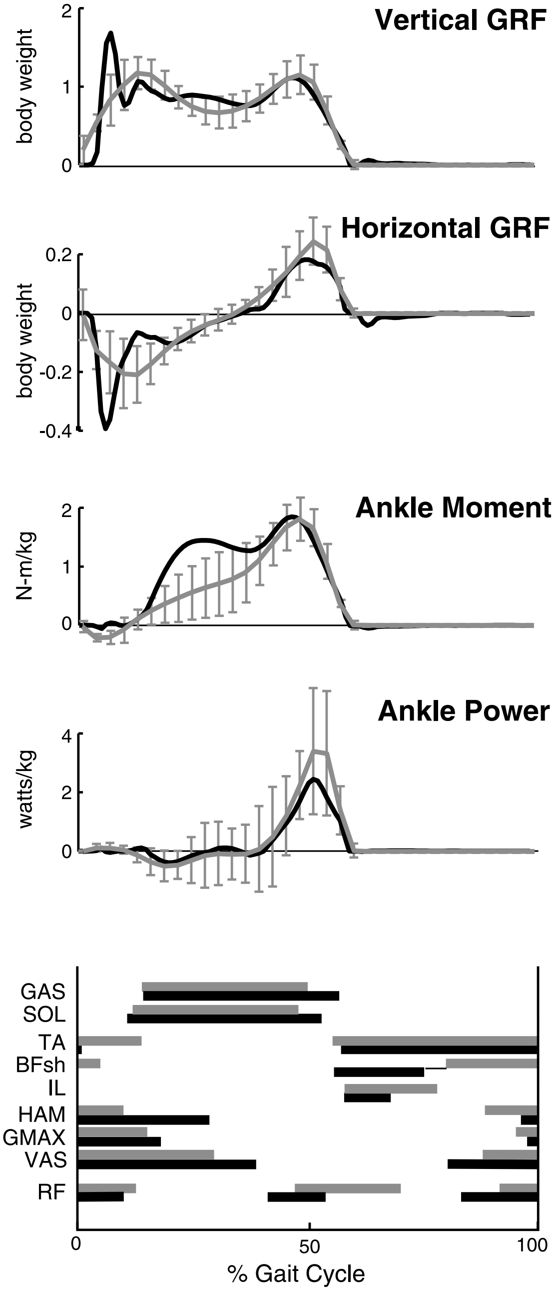

Fig. 1. Comparison of kinetics and muscle excitation phasing from a dynamical simulation with the measurements from healthy young adult subjects walking at 1.5 m/s. Black lines (kinetic trajectories) and black horizontal bars (muscle excitation phasing) from the simulation. Gray lines (average kinetic trajectories ±1 SD) and gray horizontal bars (average EMG phasing) from subjects. Simulation data from Ref. [82]. Ground reaction forces from subject measurements [161]. Ankle moment and power computed from inverse dynamics by [82] using subjects' measurements of ground reaction forces and kinematics [161]. Forces normalized by body weight; moment and power by body mass. EMG phasing from Ref. [4]. Notice the good agreement in general between the simulation and the kinetics/EMGs of the subjects. When differences exist, muscle contributions to segmental energetics and accelerations are qualitatively unchanged [82]. Ground reaction forces and ankle kinetics adapted from Ref. [82].

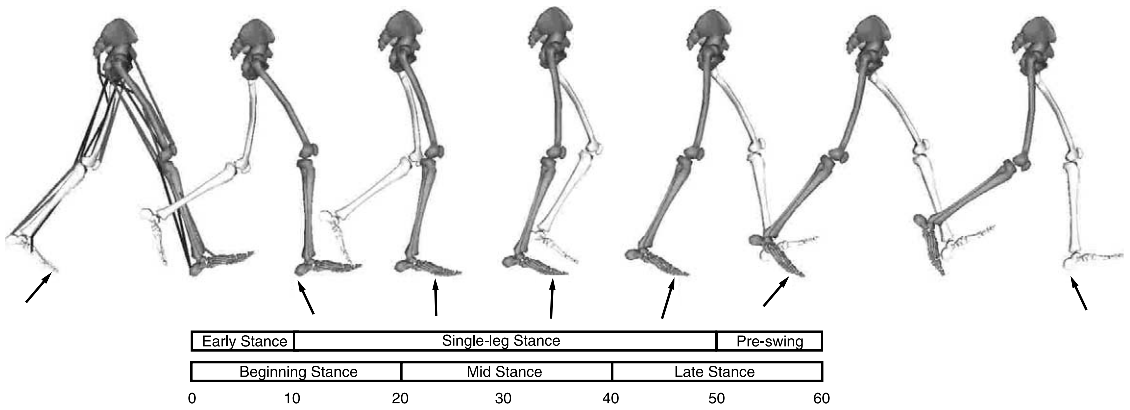

Fig. 2. Snapshots of the kinematics of the leg and pelvis of a muscle-based dynamical simulation of walking at 1.5 m/s [82]. Simulation driven by 14 muscles (left most snapshot) organized into nine groups (see Fig. 4; Part I [1]) with muscles in each group excited by the same excitation signal. Ground reaction force at each frame is the summed force from the 30 ground contact elements used to model the foot/ground interaction [115]. 'Early Stance', 'Single-leg Stance', and 'Pre-swing' gait cycle durations are defined according to convention. 'Beginning Stance' (0–20%), 'Mid Stance' (20–40%), and 'Late Stance' (40–60%) indicate approximate regions referred to in text when discussing muscle contributions to segmental energetics. Regions are shown in percent of the gait cycle. Adapted from Ref. [82].

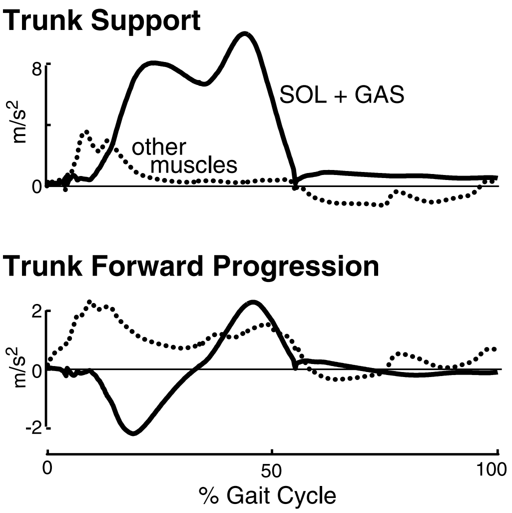

Fig. 3. Support and forward progression of the trunk provided by the plantar flexors (*SOL+GAS*; solid lines) and the other muscles (*other muscles*; dotted lines) during the gait cycle. Contribution to support and forward progression of the trunk by a muscle force is defined to be its contribution to the upward and forward acceleration of the trunk center-of-mass, respectively. VAS and GMAX contribute to support and progression in beginning of stance; RF to progression in late stance [83]. Adapted from Ref. [82].

6.4. Synergism of the uni- and biarticular plantar flexors in mid stance

In mid stance, forward motion of the trunk and leg seem to be maintained because of the synergistic action of the uni- (SOL) and biarticular (GAS) plantar flexors. SOL and GAS have been found to power the trunk and leg oppositely (Fig. 4) [82]. Together, SOL and GAS deliver little net power to the trunk. Energy flow to the trunk in the vertical direction or in anterior–posterior tilting is low because vertical and tilting trunk motion is slow [23,35]. SOL and GAS also have opposite effects on the shank and thigh (Fig. 4). No power is delivered to the foot because it is hardly moving. But SOL and GAS support the trunk (see above), and also the leg because they produce upward ankle and knee intersegmental forces. Dynamical simulations seem to show, therefore, that the summed effect of the plantar flexors in mid stance is to ensure the body is supported so it can maintain its forward motion, consistent with the notion that the plantar flexors produce a 'controlled roll-off' by preventing the leg from collapsing [32,82,118,119].

The synergism of the uni- and biarticular plantar flexors in mid stance to maintain body support and the forward motion of the trunk and leg is also consistent with inverted-pendulum-like ballistic walking [63,69,70]. First, the overall energy of the trunk is about constant in mid stance with exchange in its potential and kinetic energy occurring [120,121]. Second, the energy of the leg is also about constant then [120]. Finally, SOL and GAS muscle fibers are acting nearly isometrically [122] as they execute their segmental energetic functions in mid stance; thus their metabolic energy consumption is expected to be low [123]. Thus, the synergism of SOL and GAS in mid stance to maintain forward motion and support of the body occurs with minimal metabolic energy expenditure, as expected in ballistic-like walking.

6.5. Uni- and biarticular plantar flexors contribute differently to trunk energetics in late stance

Though the uni- (SOL) and biarticular (GAS) plantar flexors both shorten in much of late stance [122], and provide support and forward progression then (see above), they execute different mechanical segmental energetic functions [82]. The finding from muscle-based simulations that the plantar flexors produce nearly all

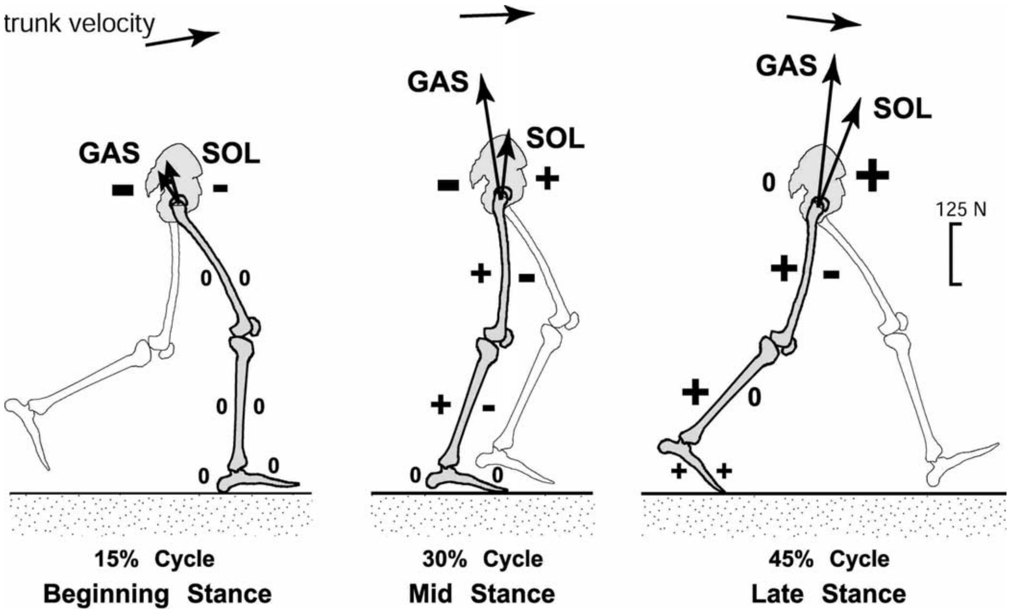

Fig. 4. Ankle plantar flexor (SOL and GAS) contributions to the hip intersegmental force and energy flow in the trunk and leg in the beginning of stance (~ 15% gait cycle), middle of stance (‘mid stance’; ~ 30% cycle), and late stance (~ 45% cycle) computed from a dynamical simulation. Stance-leg is ‘gray-filled.’ Contribution of each muscle force to the ground reaction force was computed first (see text). The net contribution of a muscle force to the hip intersegmental force (shown in figure) includes the contribution from its effect on the ground reaction force. GAS and SOL together provide some support of the trunk but retard its forward progression in the beginning of stance, provide support near mid stance, and provide progression as well as support in late stance. Notice that the contribution to the hip intersegmental force from GAS is directed more posteriorly than SOL throughout stance; i.e. relative to SOL, GAS decelerates trunk forward motion in beginning of stance over a longer region of the gait cycle and accelerates trunk forward motion in late stance over a shorter region. Velocity of the trunk is shown by the arrow at top. Trunk moves upward before mid stance, then downwards afterward in late stance [23,35]. Dot product of each induced hip force vector with the trunk velocity vector gives the net contribution by the muscle to the translational power of the trunk. Net energy flow to each segment is denoted by a ‘+’ or ‘−’ sign. Direction of SOL and GAS induced energy flow is to a large extent opposite, causing them to have a synergistic function in mid stance but different functions in late stance (see text). Bar at right to scale intersegmental forces equals 25% weight of trunk (HAT) segment (wt. = 500 N). Contributions computed from data generated from simulation in Ref. [82].

the musculotendon work in late stance is consistent with suggestions from net joint power analyses using inverse dynamics [124–126]. However, muscle-based simulations have revealed the different segmental energetic functions of the uni- and biarticular plantar flexors.

The energy produced by GAS in late stance is delivered to the leg to accelerate it forward (Fig. 4 GAS) [82]. GAS does not deliver any net energy to the trunk because the decrease in vertical energy of the trunk from its deceleration of the downward motion of the trunk equals the increase in horizontal energy of the trunk from its acceleration of the forward motion of the trunk. GAS instead accelerates the thigh, shank and foot forward, which is probably critical to leg progression, because the leg has high forward momentum and kinetic energy at that point [125,127].

In contrast, the energy produced by SOL in late stance is delivered to the trunk to accelerate it forward (Fig. 4, SOL) [82]. But the increase in trunk energy caused by SOL is more than the energy produced by SOL because SOL decelerates the thigh and shank and redistributes the energy to the trunk.

Thus, in late stance, SOL and GAS have different net energetic effects on the trunk, thigh and shank. Rather than these effects working together inseparably to execute one identifiable task, SOL and GAS instead execute distinct subtasks; SOL contributes to trunk forward progression and GAS to swing initiation, which is forward progression of the leg [82]. SOL causes the forward kinetic energy of the trunk to increase substantially. GAS causes the energy of the thigh and shank to increase. On the other hand, SOL and GAS increase the energy of the foot similarly because both muscles accelerate the foot into plantar flexion. Previous non-muscle based analyses suggested that the net effect of the combined plantar flexors is either for an active push-off [81,126] or to initiate swing [124,128]. The muscle-based simulations showed that the plantar flexors indeed contribute significantly to both effects, but their contribution to push-off arises from SOL, not GAS, and their contribution to swing initiation from GAS, not SOL [82].

6.6. *Elastic storage of energy by the uni- and biarticular plantar flexors*

Simulations of walking driven by individual muscles connected to tendons have shown that both the uni-

(SOL) and biarticular (GAS) plantar flexors store energy in the Achilles tendon and their aponeuroses in the first part of single-leg stance and release it in late stance when the muscles shorten or undergo relaxation [82,94]. The storage of energy in the series elastic structures of the human plantar flexors had been shown long ago by Hof et al. [122,129–134] who used EMG-to-force processing methods with a triceps surae musculotendon model. Energy storage has also been shown to occur in the plantar flexors of the cat during locomotion [135].

A significant amount of energy storage occurs during early single-leg stance because the net effect of the hip intersegmental forces on the trunk caused by SOL and GAS, especially GAS, is to reduce the energy of the trunk in early single-leg stance with little energy redistributed to the leg (Fig. 4, GAS) [82]. Because the Achilles tendon and the aponeuroses of SOL and GAS are relatively long [114], a considerable amount of the absorbed energy is stored in these elastic structures rather than dissipated in the muscle fibers [94,122].

The stored elastic energy is released in late stance to augment the energy produced by the contractile machinery [94,122] in order to provide trunk forward progression (SOL) or swing initiation (GAS) [82]. The release of the stored elastic energy in late stance, and especially in pre-swing which is double-leg stance near toe-off, is a most effective time to impart energy to the trunk [76]. It is important to recognize that SOL and GAS do not have to be excited throughout pre-swing [118] but would only have to be undergoing relaxation [32] in order for the stored elastic energy to be released.

Storage and elastic energy in the beginning of single-leg stance and its release in late stance is, therefore, an efficient mechanism used by SOL and GAS to provide forward progression of the trunk (SOL) and the leg (GAS) while they support the trunk. In addition, their isometric activity in mid stance provides continuity, ensuring support of the whole body so it can maintain its forward motion through mid stance (see above).

6.7. Contributions of quadriceps and hip extensors to trunk support and forward progression

Muscles other than the ankle plantar flexors also contribute to support and forward progression in the beginning of stance, and to forward progression in late stance (Fig. 3) [82]. The uniarticular quadriceps muscles (vasti group, VAS) and the uniarticular hip extensors (e.g. gluteus maximus, GMAX) are the principal muscle contributors to support and forward progression in the beginning of stance [83]. The biarticular quadriceps muscle (rectus femoris, RF) is a significant contributor to forward progression in late stance [83].

The quadriceps accelerate the trunk in the beginning of stance even though it is well accepted that they brake/decelerate knee flexion then by acting eccentrically [4,32,35]. The uniarticular quadriceps muscles (VAS) are much more effective than the biarticular quadriceps muscle (RF) [83], consistent with observation that RF activity is sometimes absent in the beginning of stance [32]. Deceleration of the knee by the quadriceps is consistent with the decrease in energy of the leg then [35,120,124,128]. Importantly, the energy gain of the trunk caused by the quadriceps is approximately the same as the energy dissipated by their muscle fibers, with very little being stored in series elastic structures, such as the quadriceps tendon [83]. Thus, the role of quadriceps (VAS particularly) to act as an accelerator of the trunk seems to be comparable to its role to act as a brake of the leg [83].

The important function of VAS to accelerate the trunk and brake the leg in the beginning of stance (Fig. 5) occurs irrespective of whether it is lengthening, shortening, or neither. In early stance, soon after heel contact, the knee flexes and VAS lengthens; so VAS acts eccentrically. Afterwards the knee briefly neither flexes nor extends and VAS neither lengthens or shortens. Then the knee extends and VAS shortens [23,32]. Nevertheless, VAS produces an upward and forward directed hip intersegmental force throughout the begin-

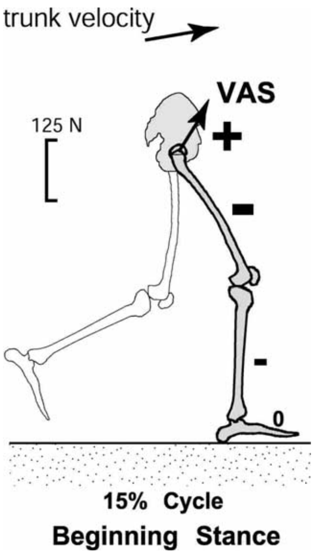

Fig. 5. Quadriceps contribution to the hip intersegmental force and energy flow in the trunk and leg in the beginning of stance (~ 15% gait cycle) computed from a dynamical simulation. Stance-leg is 'gray-filled.' The induced hip intersegmental force by the uniarticular quadriceps muscles (VAS) provides trunk support and progression in beginning of stance. Though VAS decelerates the leg ('−' signs), it also accelerates the trunk ('+'). The biarticular quadriceps muscle (RF), if active, redistributes energy qualitatively similarly (not shown; see text). The uniarticular hip extensors produce a similarly directed but much smaller hip intersegmental force and also redistribute energy among the segments similarly (not shown). Velocity of the trunk is shown by the arrow at top. Bar at right to scale hip intersegmental force equals ~ 25% weight of trunk segment. Contributions computed from data generated from simulation in Ref. [82].

ning of stance (Fig. 5) [83]. Thus, the critical role being executed by VAS is generation of force, since the direction of segmental energy changes resulting from its force generation are identical irrespective of whether it is acting eccentrically, concentrically, or isometrically.

In late stance, RF, which is active [32] and lengthening, acts to accelerate into extension both the knee and the hip [83]. These conclusions based on analyses of simulations are consistent with the notion that RF activity increases when the knee flexes too fast [32] but are inconsistent with the notion that RF 'lifts the leg to swing it forward' [4] or 'flexes the hip to accelerate the leg into swing' [136]. Also, the redistribution of energy from the leg to the trunk by RF muscle force (Fig. 6) is comparable to the energy dissipated by RF muscle fibers [83]. Therefore, RF in late stance acts energetically similar to itself in early stance, if active then, and to VAS in early stance. Thus, RF in late stance is antagonistic to the contribution by GAS to initiate swing, yet agonistic to the contribution by SOL to accelerate the trunk forward.

6.8. *Muscle compensatory mechanism in late stance*

SOL, GAS and RF may work synergistically in late stance to deliver energy to the trunk for forward progression because GAS produces mechanical energy and delivers it to the leg and both SOL and RF redistribute energy from the leg to the trunk. It may seem, therefore, that a decrease in SOL activity could be compensated by an increase in RF activity. However, such compensation by RF alone would be insufficient because SOL shortens and produces work output, which also is used to accelerate the trunk forward. GAS, which also acts concentrically, must increase its activity as well because RF acts eccentrically. Increased GAS activity would deliver more energy to the leg, which could be redistributed to the trunk through the increase in RF activity. However, this compensatory action by RF and GAS for a reduction in SOL activity would be inefficient because RF would dissipate some mechanical energy from the leg in the process, whereas SOL would not. In addition to this synergistic role with SOL and RF, GAS contributes to swing initiation (see above).

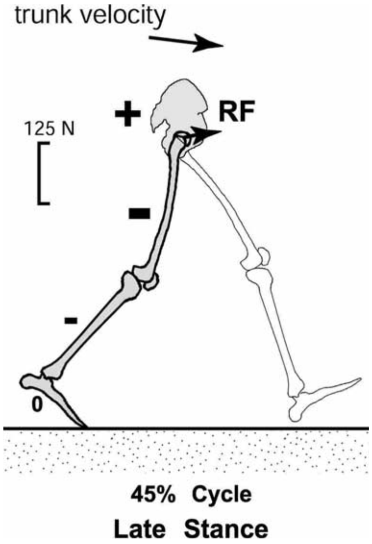

Fig. 6. Contribution of the biarticular quadriceps muscle (RF) to the hip intersegmental force and energy flow in the trunk and leg in late stance (~45% cycle). RF, acting eccentrically, causes qualitatively similar redistribution of segmental energy in late stance as VAS does in beginning of stance, i.e. deceleration of the leg and acceleration of the trunk; thus RF assists trunk progression. Velocity of the trunk is shown by the arrow at top. Bar at right to scale hip intersegmental force equals ~25% weight of trunk segment. Contributions computed from data generated from simulation in Ref. [82].

6.9. *Mechanical energy production by muscles over the gait cycle*

It is clear from muscle-based simulations and from net joint power analyses [124,125,127,128] that muscles work together throughout the gait cycle to redistribute much energy among the body segments to support and propel them forward. However, muscles not only must generate force to cause segmental energy redistribution but also must produce work output to replenish the energy lost during impact with the ground, the mechanical energy lost when muscles act eccentrically, and the energy lost via viscous damping in passive joint stiffnesses.

Sagittal-plane simulations suggest that hamstrings (HAM), the plantar flexors (SOL and GAS), and the uniarticular hip extensors (GMAX) produce most of the mechanical energy over the cycle [83]. However, GAS and HAM do not deliver the energy they produce directly to the trunk by accelerating it. Instead, they deliver their energy to the leg(s). Thus, other muscles are either co-excited and redistribute the energy to the trunk by acting synergistically, such as coexcitation of RF and SOL with GAS in late stance, or other muscles are excited later and redistribute the energy to the trunk then. Analysis of these simulations show that the simulated work produced by all muscles per body mass and unit distance traveled is, not unexpectedly, slightly higher than the minimum work predicted from muscles when the body is considered to be an inverted pendulum where its center-of-mass exchanges gravitational potential energy with kinetic energy (i.e. 0.3 J/kg/m, [137]).

6.10. *Sensitivity of muscle coordination principles to simulation data*

The simulation-derived coordination principles reviewed here are believed to be rather robust to the parameters characterizing the dynamical model of the body because the simulation data are compatible with measured walking kinematics, kinetics, and EMGs. Furthermore, the dynamical model of the body used in the simulations [46–48], or its precursors, has more than a decade-long history of usage in studies of human legged motor tasks [94,138–144].

Nevertheless, it is important to realize that the kinetic quantities being computed from the simulation data to infer muscle function are unmeasurable, such as muscle-induced accelerations and segmental power. Thus, the confidence in the conclusions must be based on sensitivity studies, which show how much variation in parameter values or model structure can exist for the conclusions of muscle function to remain unaltered. As with the continued collection of experimental gait data and analyses to address a specific scientific hypothesis, the study of the sensitivity of simulated gait data is also an ongoing process.

Though we feel that the function of muscles described above is rather robust to model parameters and structure, future sensitivity studies will undoubtedly define better the limitations of these descriptions. Our sensitivity studies so far indicate that the experimental determination of the moment arms for the biarticular muscles is a particular area where more precise data would be useful. Similarly, a more detailed model of ground contact and more complete understanding of its relationship to muscle forces will need to be developed. In studies of pathological gait, we expect that simulations will have to be generated using 3D musculo-skeletal models, even to understand trunk forward progression and vertical support, due to the large frontal and transverse plane motions.

6.11. *Clinical applications*

We believe that static musculo-skeletal models can be a productive first step to enhance the understanding of locomotor performance in not only healthy individuals but also individuals with impairments. Musculo-skeletal models have been used in conjunction with kinematic measurements to study the causes of crouch gait and to estimate the lengths of muscles being considered for surgery [145,146]. Musculo-skeletal models have also been used to examine the causes of excessive internal rotation of the hip [147–149] and to emulate the biomechanical consequences of surgeries performed to improve locomotor performance, including tendon lengthenings [150,151], tendon transfers [152], osteotomies [153–155] and joint replacements [156,157].

While these static musculo-skeletal models provide some insights, gait is a dynamic motor task and muscle-based dynamical simulations are needed to better understand the function of muscles and the cause of gait abnormalities. Using a muscle-actuated dynamical model to simulate the swing phase, Piazza and Delp [158] concluded that hyperactivity in RF in swing, and presumably in pre-swing as well, has the potential to produce stiff-leg gait. Riley and Kerrigan [159] reached a similar conclusion analyzing a forward dynamic model of early swing in subjects with post-stroke unilateral stiff-knee gait. However, stance-phase factors that decreased knee flexion velocity at toe-off also were shown by Piazza and Delp to be possible contributors to stiff-knee gait. In fact, above normal force generation in SOL in late stance could lead to stiff-knee gait because dynamical simulations [82] have shown that SOL acts powerfully then to accelerate the knee into extension. It should be emphasized that the identification of the muscle coordination mechanisms responsible for pathological gait is quite challenging, given the difficulty in understanding muscle coordination of gait in even healthy individuals.

It is even more challenging to develop patient-specific dynamical simulations of walking. The first challenge is to determine the limitation of simulations based on generic musculo-skeletal models. For example, what are the limitations of using a generic musculo-skeletal model to generate patient-specific simulations that replicate their individual kinematics, kinetics, and muscle activity patterns? Even for a class of patients, the generic musculo-skeletal model may have to be different from even an age-matched model of healthy individuals. Limb deformities may necessitate changes in the model describing the joint kinematics and the muscle moment arms [160]. And if the simulations using this generic model show high sensitivity to the specific musculo-skeletal parameters describing these deformities, then methods would have to be developed to measure these parameters in individual patients. Another significant limitation is the lack of uniqueness of the simulation to reproduce patient-specific kinematics, kinetics, and muscle activity, even when the musculo-skeletal model is patient-specific. The reason fundamentally is that the repertoire of measurements possible in subjects is limited, and often even more restrictive in patients. We believe that the number of muscles from which reliable, or reproducible activity can be measured will ultimately set the bound on our confidence in the interpretation of the simulation data.

7. Concluding remarks

In this Part II, we have reviewed how dynamical simulations derived from simple and complex models have been used to study many facets of bipedal walking. At the one extreme, simulations derived from multi-segmental models without muscles have shown that the body has a natural propensity for locomotion once it is moving because of the intersegmental dynamics of the legs and trunk. At the other extreme, simulations derived from neuro-musculo-skeletal models, which incorporate concepts of neural locomotion to excite muscles in multisegmented legs, have demonstrated how stable gait can be produced. Because computational power is expected to continue to increase, it appears that dynamical simulations of walking will become increasingly used to study muscle coordination of gait.

Analysis of simulations driven by actuators with muscle-like properties are now beginning to provide insight into muscle coordination mechanisms. As reviewed above, simulations have shown that the visco-elastic properties of muscles tend to stabilize gait. Simulations have also shown that the force generated by a muscle can cause significant energy flow among the segments through its contributions to the joint intersegmental forces.

The importance of muscle force generation to the interchange of segmental energy is consistent with ballistic and ‘passive’ walking concepts. Muscles produce force to support the body, which permits the body segments to redistribute their mechanical energy. And the directionality of the segmental energy redistribution caused by a muscle force occurs regardless of whether the muscle is acting eccentrically, concentrically, or isometrically. Of course, the physiological cost for the energy redistribution is affected by whether the muscle is lengthening, shortening, or at a constant length.

Muscles have to participate in the energy exchange among the body segments and the environment because inertial and gravitational forces alone are insufficient to achieve the task goals, evidenced by the summed mechanical energy state of the system being not constant over the locomotor cycle. Simulations suggest that the ankle plantar flexors (SOL, GAS) and the uni- and biarticular hip extensors (GMAX, HAM) dominate work output over the gait cycle [83]. These muscles, being active in late stance and the beginning of stance, are therefore restoring energy to the body near double-support, a most effective time to utilize the passive locomotion properties of the body.

The deduction of the role of individual muscles in gait by analyzing simulations is arising largely from computations of the instantaneous contributions of muscle force generation to the acceleration and power of the segments. However, these muscle-induced accelerations over the gait cycle are a series of snapshots in time and do not necessarily account for the cumulative effects the past muscle force trajectories have had on system behavior (see Part I [1]). Attempts to decompose the ground reaction force into contributions arising from individual muscles is a step toward accounting for the cumulative effects. Nevertheless, only through the development of methods that can better account for the effects of the past system behavior on the acceleration and movement of the segments will we know the limitations of our understanding of muscle coordination derived from current simulation analyses.

Dynamical simulations of gait have so far emphasized the basic principles of coordination of the segments and the basic role of individual muscles. It is likely that most immediate future simulations will continue along this path. Simulations will be generated to replicate gait data obtained from subject populations, such as the healthy elderly or children, to understand how coordination in these populations differs from those in young healthy adults. Nevertheless, we believe that effort must be expended to develop methods to create dynamical simulations of individual subjects or patients. The reason is that treatment success in individual patients will rise with more patient-specific data. Patient-specific simulations, by their ability to produce unmeasurable patient-specific data consistent with all measured patient data, offer, therefore, hope in achieving the goal of better treatment.

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