Published in final edited form as:

J Biomech. 2009 August 7; 42(11): 1786–1789. doi:10.1016/j.jbiomech.2009.04.029.

Author's Response to Comment on “Contributions of the individual ankle plantar flexors to support, forward progression and swing initiation during walking” (Neptune et al., 2001) and “Muscle mechanical work requirements during normal walking: The energetic cost of raising the body's center-of-mass is significant” (Neptune et al., 2004)

Richard R. Neptune¹, Felix E. Zajac², and Steven A. Kautz^{3,4,5}

¹ Department of Mechanical Engineering, The University of Texas at Austin, Austin, TX, USA

² Department of Mechanical Engineering and Department of Orthopedic Surgery, Stanford University, Stanford, CA, USA

³ Brain Rehabilitation Research Center, Malcom Randall VA Medical Center, Gainesville, FL, USA

⁴ Department of Physical Therapy, University of Florida, Gainesville, FL, USA

⁵ Brooks Center for Rehabilitation Studies, University of Florida, Gainesville, FL, USA

We appreciate the opportunity to respond to Kuo and Donelan's letter (Kuo and Donelan, 2009) and hope our response encourages additional dialogue regarding muscle mechanical work and metabolic cost during walking. Below we clarify that much of their critique is based on misguided assumptions regarding our forward dynamics simulations. Further, they inappropriately differentiate “experimental data” (e.g., net joint work -- which is actually derived from linked-segment models) from modeling data and erroneously treat joint and center-of-mass (COM) work as a gold standard for inferring muscle work. Finally, we provide more evidence in support of our conclusion that much muscle work is required in the beginning of single-support, most likely to raise the COM.

Misguided assumptions regarding our forward dynamics simulations and conclusions

Many of their analyses are invalid because they either improperly combine data from two simulations with different results or make incorrect assumptions regarding the simulation data. First, net joint work from the 2001 simulation (overground walking using block patterns for the muscle excitations) cannot be equated to the muscle work in the 2004 simulation (treadmill walking using EMGs for the muscle excitations with no center-of-pressure data available to compute net joint powers). Therefore their statement “one result from the Model

Please address correspondence to: Richard R. Neptune, Ph.D., Department of Mechanical Engineering, University of Texas at Austin, 1 University Station C2200, Austin, TX 78712 USA, rneptune@mail.utexas.edu.

Publisher's Disclaimer: This is a PDF file of an unedited manuscript that has been accepted for publication. As a service to our customers we are providing this early version of the manuscript. The manuscript will undergo copyediting, typesetting, and review of the resulting proof before it is published in its final citable form. Please note that during the production process errors may be discovered which could affect the content, and all legal disclaimers that apply to the journal pertain.

does not agree with another” is inappropriate. Second, our 2004 simulation does not violate physical principles. Their calculation of an average bilateral vertical GRF = 1.1 bodyweight over the gait cycle by assuming perfect bilateral symmetry is incorrect (actually 0.997 bodyweight in our model) because the impact peak is lower in the contralateral leg even though symmetrical muscle controls were used. Perfectly symmetrical simulations are difficult to achieve because of inherent instability of bipedal locomotion. Third, we find no error in our experimental COM work data. Kuo and Donelan provide no rationale for why our subjects could not walk with the presented COM work rate trajectory given the variability found among subjects. Further, their own COM data at 1.5 m/s has a net COM work = ~ -2.5 J (see Donelan et al., 2002, Fig. 5), which violates fundamental physical principles by their standards, though it demonstrates well the inevitable inaccuracies in calculations based on “experimental” data (i.e., computed COM work is not perfectly consistent with symmetrical, steady-state walking). Finally, our 2004 study analyzed muscle mechanical work, not metabolic energy expenditure. We only noted that our results were “consistent with others’ suggestions that a major determinant of the metabolic cost of walking is raising the COM” (Neptune et al., 2004) and did not conclude that muscle work in Region 2 “is a major contributor to the metabolic cost of human walking, with the purpose of raising the COM” (Kuo and Donelan, 2009).

Misleading differentiation between experimental and modeling data

They refer to our results as “Model” and refer to published experimental data as “Human” in an attempt to show that our Model does not replicate the “Human” well. But the “Human” data referred to (i.e., net joint power and work) are calculated from linked-segment models -- not measured. These models, as ours, require a model of the joints and the physical properties of the body segments. Thus, their inference that our results are from a model, and theirs are not, is misleading.

Inappropriate use of joint and COM work as a gold standard to infer muscle work

We acknowledge that our simulations do not perfectly reproduce net joint power and work quantities computed from inverse dynamic models. However, discrepancy may be due in part to experimental errors in kinematic and kinetic measurements (e.g., inaccurate marker registration and skin movement artifact), which would make reproduction by a kinematically consistent model virtually impossible. Further, joint moment-based quantities derived from inverse dynamics often have high residuals (e.g., Kuo, 1998) that violate physical principles (i.e., they represent non-existent forces acting on the system) due to dynamic inconsistency between the GRFs and body segment kinematics, causing net joint work quantities to be dynamically inconsistent. Error identification is difficult using inverse dynamics because no independent method is available to validate joint work quantities since energy dissipated in the musculoskeletal system and the environment is unknown. This often leads to large variability in joint powers between studies (e.g., compare the sagittal hip moment power in Winter (1991) versus Eng and Winter (1995)). Had Kuo and Donelan used Eng and Winter (1995), which are data for a walking speed closer to ours, instead of Winter (1991) for their “Human” data, they would have found that our joint work in Region 2 was actually lower, not higher, than the “Human.” Thus, we believe Kuo and Donelan should use caution when stating that “experimental” data (measurements; joint quantities computed from inverse dynamics) must serve as the arbiter.

Even more problematic than the unacknowledged shortcomings in the inverse-dynamics-derived calculations of joint work is Kuo and Donelan’s inference of muscle work from analyses of net joint work and COM work. Previous analyses (Neptune and van den Bogert, 1998; Sasaki et al., 2009) have clearly highlighted the inability of these measures to accurately predict

muscle work. COM work and joint work are computed net measures that cannot partition the amount of positive and negative muscle work performed, primarily because of muscle co-contraction and multiarticular muscles (Sasaki et al., 2009). In contrast, muscle-driven forward dynamics simulations use fundamental laws of physics to ensure kinematic and dynamic consistency at each point in the gait cycle while accounting for simultaneous positive and negative work. Thus net joint and COM work should not be used as the gold standard to estimate muscle work.

Muscle work in early single leg support (Region 2) is significant

We believe that the crux of the issue is whether much muscle work is produced in Region 2, as we maintain. If so, step-to-step transition theory (Donelan et al., 2002; Kuo et al., 2005) is refuted, at least in its totality as it purports that very little muscle work is needed after the heel-strike transition in Region 2. Kuo and Donelan suggest little muscle work occurs in Region 2 based solely on an analysis of net joint work. However, inferences of muscle work based on net joint work are fundamentally flawed (Sasaki et al., 2009), as well as at odds with EMG activity during the first half of stance (e.g., Perry, 1992), especially when electromechanical delay of force generation (e.g., Norman and Komi, 1979) is considered.

Our recent simulations, which track measurements and computed data even better (e.g., the GRFs; COM work; Fig. 1), still show musculotendon work is required in Region 2 (Fig. 2). These simulation results are consistent with Sawicki and Ferris (2009) who, using powered ankle exoskeletons to augment ankle plantarflexor power in push off (Region 4, where the step-to-step transition theory suggests the majority of positive muscle work occurs), found that the plantarflexor muscles consume only 18% of the total net metabolic power at the preferred speed/step length combination, with muscle work at the proximal joints elsewhere in the gait cycle other than in Region 4 (i.e. hip and knee; Region 2 perhaps?) accounting for the remaining 82% of the metabolic cost of walking. Therefore, we stand by our conclusion that significant muscle work is produced in Region 2.

Muscle work in Region 2 raises the COM

We stand by this suggestion, which is counter to the step-to-step transition theory assertion that the COM rises because of a passive exchange of kinetic and potential energy. We and others (e.g., Anderson and Pandy, 2003; Liu et al., 2006) have performed analyses that show muscle work is indeed required in Region 2, even though the decrease in body kinetic energy approximates the increase in potential energy then. Muscle force is required to accelerate the COM vertically because the hip and knee joints are flexed in Region 2. Without muscle force (work), the joints would collapse rather than extend, and the trunk would not rise. Kuo and Donelan argue that, should musculotendon work be expended, the vertical GRF would be expected to be higher than body weight in Region 2 when the COM rises, which it is not. Our simulations show that without muscle forces the body would collapse and the vertical GRF would be even lower. The vertical GRF is less than body weight in Region 2, and yet the COM is accelerated upwards, because (i) the contribution of gravity to the vertical GRF is less than 1/2 body weight, due to the flexed body segments, (ii) muscle force contributions exceed gravity's, and (iii) muscle and gravity's summed contribution is still <1 (Anderson and Pandy, 2003) (cf. stiff leg gait, or standing upright, where the joints are hardly flexed and gravity's contribution to the GRF is ~ 1 body weight). Thus muscle forces extend the hip and knee, and produce work, even though the vertical GRF is less than 1 body weight. We are skeptical that passive body mechanics can achieve similar behavior, given the observed body segmental geometry and kinematic trajectories. We believe that our interpretation is based on sound, objective biomechanical data with explicit definitions of segmental and COM vertical

acceleration, and individual muscle and gravity contributions, and not overly subjective (Kuo and Donelan, 2009).

In short, we believe muscle based models must be used to understand muscle coordination of walking and, by definition, non-muscle based models cannot be used. However, non-muscle based models of walking (e.g., Donelan et al., 2002; Kuo et al., 2005) can be useful to the understanding of basic mechanical principles operating across animal species and sometimes human behavior, such as metabolic energy expenditure (Zajac et al., 2002). But it should be recognized that metabolic energy expenditure is not a within-gait-cycle measurement but rather an average across many gait cycles. We believe (i) our simulations of muscle coordination of walking do replicate the within gait cycle data well, as well as muscle architecture and dynamics, and musculoskeletal geometry, which non-muscle based models ignore completely, (ii) our conclusion that significant muscle work is generated in the beginning of single stance is robust, and (iii) our suggestion that this work is used to raise the COM is reasonable.

Acknowledgments

This work was supported in part by NIH grants RO1 HD46820 and RO1 NS55380.

References

- Anderson FC, Pandy MG. Individual muscle contributions to support in normal walking. *Gait Posture* 2003;17(2):159–69. [PubMed: 12633777]
- Donelan JM, Kram R, Kuo AD. Simultaneous positive and negative external mechanical work in human walking. *J Biomech* 2002;35(1):117–24. [PubMed: 11747890]
- Eng JJ, Winter DA. Kinetic analysis of the lower limbs during walking: what information can be gained from a three-dimensional model? *J Biomech* 1995;28(6):753–8. [PubMed: 7601875]
- Kuo AD. A least-squares estimation approach to improving the precision of inverse dynamics computations. *J Biomech Eng* 1998;120(1):148–59. [PubMed: 9675694]
- Kuo AD, Donelan JM. Comment on “Contributions of the individual ankle plantar flexors to support, forward progression and swing initiation during walking” (Neptune et al., 2001) and “Muscle mechanical work requirements during normal walking: The energetic cost of raising the body's center-of-mass is significant” (Neptune et al., 2004). *J Biomech*. 2009 submitted.
- Kuo AD, Donelan JM, Ruina A. Energetic consequences of walking like an inverted pendulum: step-to-step transitions. *Exerc Sport Sci Rev* 2005;33(2):88–97. [PubMed: 15821430]
- Liu MQ, Anderson FC, Pandy MG, Delp SL. Muscles that support the body also modulate forward progression during walking. *J Biomech* 2006;39(14):2623–30. [PubMed: 16216251]
- McGowan CP, Kram R, Neptune RR. Modulation of leg muscle function in response to altered demand for body support and forward propulsion during walking. *J Biomech*. 2009 in press.
- Neptune RR, van den Bogert AJ. Standard mechanical energy analyses do not correlate with muscle work in cycling. *J Biomech* 1998;31(3):239–45. [PubMed: 9645538]
- Neptune RR, Zajac FE, Kautz SA. Muscle mechanical work requirements during normal walking: the energetic cost of raising the body's center-of-mass is significant. *J Biomech* 2004;37(6):817–25. [PubMed: 15111069]
- Norman RW, Komi PV. Electromechanical delay in skeletal muscle under normal movement conditions. *Acta Physiol Scand* 1979;106(3):241–8. [PubMed: 506761]
- Perry, J. *Gait Analysis: Normal and Pathological Function*. Thorofare, NJ: Slack, Inc.; 1992.
- Sasaki K, Neptune RR, Kautz SA. The relationships between muscle, external, internal and joint mechanical work during normal walking. *J Exp Biol* 2009;212(Pt 5):738–44. [PubMed: 19218526]
- Sawicki GS, Ferris DP. Powered ankle exoskeletons reveal the metabolic cost of plantar flexor mechanical work during walking with longer steps at constant step frequency. *J Exp Biol* 2009;212(Pt 1):21–31. [PubMed: 19088207]

- Winter, DA. Biomechanics and Motor Control of Human Gait: Normal, Elderly and Pathological. Waterloo, Ontario: Waterloo Biomechanics; 1991.
- Zajac FE, Neptune RR, Kautz SA. Biomechanics and muscle coordination of human walking. Part I: introduction to concepts, power transfer, dynamics and simulations. *Gait Posture* 2002;16(3):215–32. [PubMed: 12443946]

Figure 1.

Comparison between measurements and inverse-dynamics-derived data (“experimental data”) and corresponding forward dynamics simulation. The musculoskeletal model and experimental data collection and processing are describe in McGowan et al. (2009).

Figure 2.

Positive (Pos), negative (Neg), net (Net) and COM (External) mechanical work produced in each of the four regions by the musculotendons units and muscle fibers (contractile element). The difference in work output between the musculotendon units and the muscle fibers is due primarily to elastic energy stored and returned in the tendon (series elastic element). Note, ~ half the musculotendon work in Region 4 is due to tendon elastic energy storage and return (i.e., the difference between the musculotendon and fiber work in Region 4 is tendon work).