What Do Walking Humans Want From Mechatronics?

(Invited Presentation)

Steven H. Collins
Department of Mechanical Engineering, Robotics Institute
Carnegie Mellon University, Pittsburgh, Pennsylvania 15213-3815
biomechatronics.cit.cmu.edu, stevecollins@cmu.edu

***Abstract*—It is an exciting time to be developing robotic prostheses, exoskeletons, and gait trainers, with clever new innovations emerging at a rapid pace. But are these the droids we're looking for? It is very difficult to predict how a human will respond and adapt to forceful interactions with an electromechanical device, and many years of development are typically required before proposed designs can be tested on humans. What if we could test our ideas for device function quickly, without the overhead of designing a product-ready prototype? This might lead to faster, and more meaningful, understanding of design requirements and trade-offs for human users. We will describe a system that we have developed for rapid emulation of robotic ankle prostheses and orthoses, and present initial results from the high-throughput experiments that this technology has enabled. One set of experiments provides quantitative insights into the optimal prosthesis motor and battery size for a given user, while another set identifies the relationships between energy cost, balance, and variability during gait. Experiments with an ankle-foot orthosis demonstrate shaping of the human energy-cost landscape, revealing that least-effort drives can be harnessed to shape self-selected coordination patterns, with applications to gait rehabilitation. We think this approach will facilitate faster identification of what humans need from wearable robots, providing detailed design requirements for engineers and resulting in better assistive technologies, sooner.**

I. Learning to Help Walk

Disabilities of the lower limbs, resulting from stroke, amputation, and other neuromuscular diseases and injuries, negatively impact quality of life for millions of individuals. One branch of the emerging field of bio-mechatronics seeks to meet the mobility needs of these individuals through wearable robots; new designs of robotic prostheses, exoskeletons, and gait trainers, intended to improve locomotor performance in terms of walking speed, energy use, or balance, are being developed at an accelerating pace. However, very few of these tools have achieved their intended goals. Here we will explore methods for developing assistive devices, try to understand why progress has been slow despite great investment, and tease out implications for fertile areas of study in mechatronics.

II. Simulations, Robots, and other Models

Many assistive robotic devices are designed based on principles derived from simplified models of the task of interest. In this approach, one tries to understand the fundamental features of the behavior by re-creating it and noting optimal coordination strategies. If one then assumes that humans use the same strategies, assistance functionalities that cooperate with the human can be inferred. Such functionalities can then be embodied by a particular robotic device. Explicitly or implicitly, this approach is very common in wearable robots.

A. Push-Off Work and Amputee Energetics

The development of actively-powered prosthetic ankle joints provides a good example. Very simple models of walking [1, 2] suggest that the energy cost of bipedal locomotion is dominated by the 'step-to-step transition', the phase during which support is transferred from one leg to the next. In particular, performing positive work with the trailing leg, or 'push-off', reduces the energy dissipated by the leading leg 'collision', and thus the energy cost overall (steady walking is net energy neutral). This principle was further explored in a walking robot [3, 4] powered by ankle push-off. A charged spring mechanism allowed for large push-off power with a small motor. This robot consumed an order of magnitude less energy, per unit weight per unit distance, than any other walking robot (recently surpassed by a similar design [5]).

These models strongly suggest a pathway for assisting individuals with lower-limb amputation. Conventional, non-robotic ankle-foot prostheses use carbon-fiber leaf springs to provide some push-off, but this is much smaller than observed in the intact ankle.

Several research groups posited that additional push-off work, perhaps provided by a motor, could reduce amputee energy consumption and increase peak walking speed. One such design [6] used an energy-recycling mechanism to capture work normally dissipated during collision and return it as push-off during the subsequent transition phase. (This approach is similar to electrical energy harvesting [e.g. 7], but without conversion inefficiencies.) In experiments with non-amputee subjects, who wore special boots to simulate the effect of amputation on one limb, the increased push-off work provided by this energy-recycling foot led to a small decrease in energy cost, about 9%. However, the same device did not decrease energy cost for amputee subjects. One prosthetic ankle that uses a motor to augment push-off has demonstrated a significant improvement in walking economy [8], and is now commercially available [9], while several similar devices have not demonstrated benefits.

978-1-4673-1388-9/13/$31.00 ©2013 IEEE

B. Subtle Mechanics, Large Performance Differences

Why did various designs all based on the same fundamental mechanical principle yield such different results, none of which were as dramatic as in models of gait? Humans respond to forceful interactions with robots in complex ways, including significant adaptation, or even growth, over time. Subtle differences in device mechanics or control can therefore produce significant differences in performance.

Consider, for example, the subtle role of the arms in human walking. The arms do not contact the substrate, and comprise little mass compared to the rest of the body, so we might not guess they play an important role in gait. Yet, curiously, people tend to move them in a characteristic way as they walk. We first became interested in the potential role of the arms by accident; while refining a passive-dynamic walking robot [10], we found that arms were needed to prevent unpredictable yaw motions mid-stance. This robot had poor ground contact, so friction was not able to provide sufficient moments to alter the angular momentum about a vertical axis as required for the legs to swing past each other. We hypothesized that with improved scrubbing friction, humans would be unlikely to slip, but might require muscular effort to sustain moments along the axis of the legs. A controlled experiment [11] revealed that metabolic energy consumption was increased 12% when arm swing was voluntarily suppressed and 26% when the arms were swung opposite from normal. To put this in perspective, deciding not to swing one's arms during walking costs about the same amount of energy as can be saved with the best robotic ankle-foot prosthesis to date. A similar case can be made with respect to knee torques and effective foot roll-over shape [12].

Difficulty in predicting significant human responses to subtle changes in their mechanical or dynamical environment significantly limits the power of intuition and simple-model based principles for the design of wearable robots. Yet the connection between proposed device functionalities and intended human benefits are rarely well understood at the outset of the development process, and are often not addressed explicitly at all. Perhaps this trend relates to the tendency of roboticists (like the author) to be more interested in *how* to build mechatronic systems than *what* functionality would most benefit the human users of the device. Regardless, a better understanding of expected human response would greatly improve efficiency in the development of assistive robots. Perhaps detailed numerical models [e.g. 13, 14] will someday allow for useful prediction of human adaptation to such devices. Until such time, perhaps we can speed the testing process.

III. Universal Wearable Robot Emulators

One way of predicting human response to a wearable robot is to imitate the function of that device using a wearable haptic interface (Figure 1). Human responses to candidate assistance strategies can thereby be studied using such emulators prior to implementation in autonomous designs. The ideal emulator would have high torque, power, and closed-loop torque bandwidth, but very low mass worn by the subject. Experimental tools can leverage the advantages of a laboratory setting, for instance using tethered off-board motor and control components, to achieve higher performance with simpler designs. Only one drive system is needed for a wide variety of end-effectors, which can be lightweight to minimize interference with natural motions. Precise control of human-robot interaction torques or forces allows emulation of common mechanical elements such as springs, with sufficient closed-loop torque bandwidth. Torque control prevents interactions being dominated by robot position, which can restrict human engagement [15].

We took our inspiration from laboratory testbeds, which have often been used as versatile exploratory tools in basic research on, e.g., human neuromechanics [16, 17]. Such systems typically serve as probes, requiring only moderate mechatronic performance to gain useful insights. With improved fidelity, we propose such tools could be used to emulate specialized, wearable robots [18].

Pilot tests of walking with the robotic prosthesis testbed have demonstrated the suitability of this experimental tool for emulating a wide variety of proposed device functions under realistic conditions [19]. We measured very low torque tracking errors and found that net work production could be systematically and consistently altered across conditions. Impedance tracking is especially difficult and important in this domain. The testbed can emulate prosthesis designs with a wide range of mechanical features hypothesized to be beneficial, and can even alter these features online, e.g., to optimize device performance with an individual user. Controlled step-by-step changes could also be used to address a variety of scientific questions, allowing direct measurement of human response to systematic changes in, e.g., dynamic stability [20] or altered metabolic cost landscapes [21].

This versatility was enabled by improved mechatronic performance compared to prior torque-capable designs, particularly in terms of worn mass and closed-loop torque bandwidth. High closed-loop torque bandwidth is important for dynamic emulation during periods of rapidly-changing conditions, such as the initial contact of the foot with the ground [22], while low mass is needed to avoid affecting natural limb motions or increasing user effort [23]. Both the prosthesis and exoskeleton end-effectors had lower mass than the lightest reported designs (0.96 kg and 0.53 kg vs. 1.37 kg in [24]), yet with an order of magnitude greater bandwidth. Benchtop tests with the prosthesis testbed revealed higher closed-loop torque bandwidth than the highest open-loop bandwidth values reported for prior designs (17 Hz vs. 14 Hz in [25]), but with less than half the mass. The testbed also exhibited higher peak torque (175 N·m vs. 134 N·m in [25]) and peak power (1006 W vs. 270 W in [26]) than prior experimental results. These results also compare well with observations of the human ankle and foot. We demonstrated peak torques 50% greater than those observed during human walking (1.6 N·m·kg^{-1} [27]), device mass less than a human foot (1.5% body mass [28]), and torque

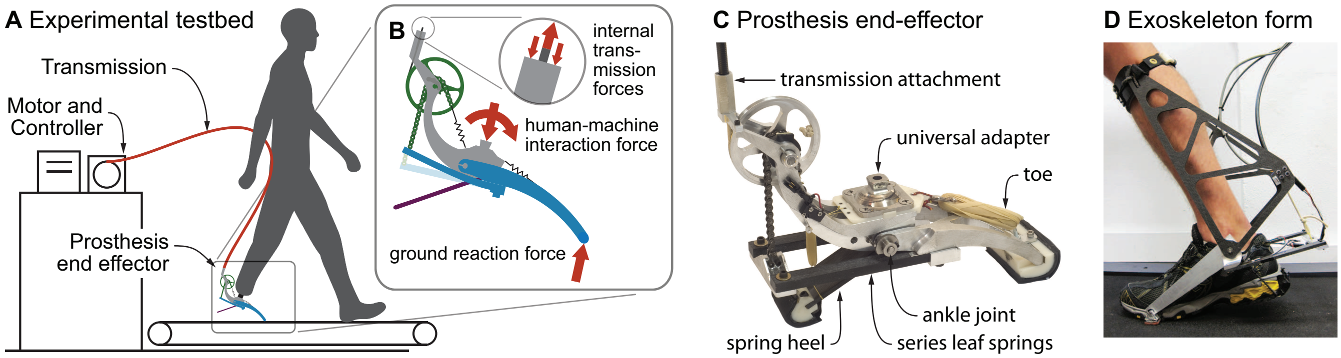

Fig. 1. Mechatronic design. **A.** The experimental testbed comprises: (1) powerful off-board motor and control hardware, (2) a flexible tether transmitting mechanical power and sensor signals, and (3) a lightweight instrumented end-effector. This division of components was chosen to maximize responsiveness and minimize end-effector mass. **B.** Free-body diagram of the end effector. Bowden cable tether forces are transmitted through a leaf spring to the toe, giving rise to ankle plantarflexion torques. **C.** Photograph of the instrumented prosthesis. A universal adapter attaches to the pylon or prosthesis simulator boot worn by the user. Fiberglass leaf springs provide series elasticity for ankle torque measurement and control. **D.** Photograph of the exoskeleton end-effector.

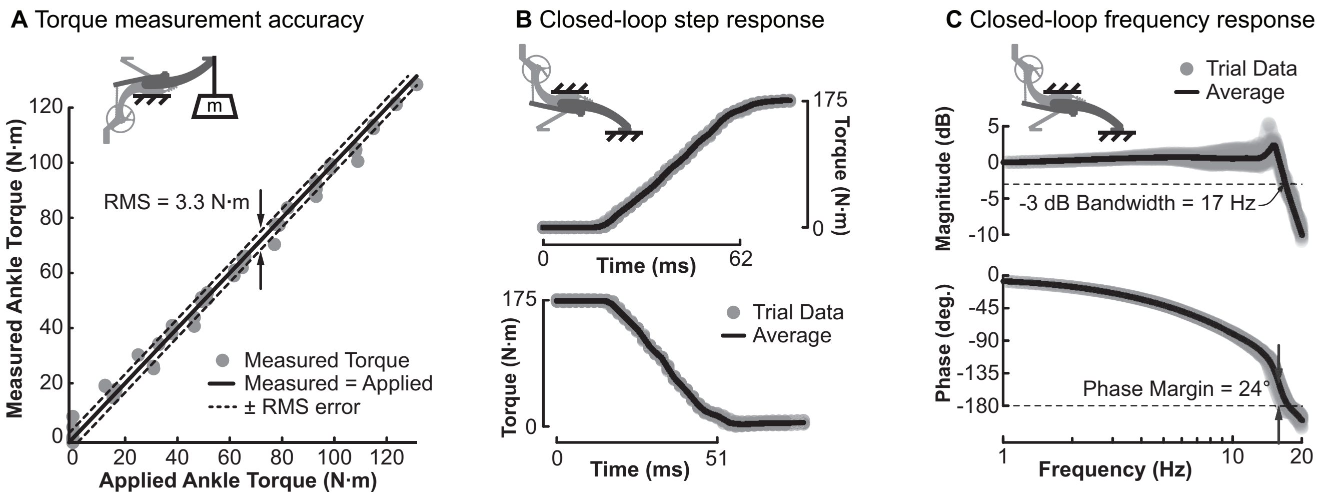

Fig. 2. Benchtop results. **A.** Torque measurement accuracy. We found RMS measurement error of 3.3 N·m and maximum error of 7.9 N·m. **B.** Closed-loop torque step response. We fixed the base and toe of the prosthesis and applied 175 N·m step changes in desired torque. Across 10 trials, we measured average 90% rise times of 0.062 s for steps up, fall times of 0.051 s for steps down, and 0% overshoot. **C.** Bode plot of frequency response under closed-loop torque control. We fixed the base and toe of the prosthesis and applied 50 N·m amplitude chirps in desired torque, then smoothed the resulting curves and averaged over 10 trials. We calculated an average -3 dB bandwidth of 17 Hz and an average phase margin of 23.6° .

bandwidth twice that of ankle muscles (6-10 Hz [29]). Some other actuators have demonstrated similar torque bandwidth, but with substantially lower peak torque and greater mass [e.g. 30–33]. Obtaining strong mechatronic performance becomes easy with this distribution of components.

A. *Initial Emulator Results*

We have used this testbed in several pilot experiments aimed at addressing questions fundamental to the benefits of ankle assistance devices. First, we systematically varied prosthetic ankle push-off work in isolation, and found a strong relationship with human energy cost. Second, we implemented discrete step-by-step ankle control strategies aimed at improving stability, based on simulation results, and measured human gait variability and energy cost. Third, we implemented a selective assistance strategy, in which desirable behavior was rewarded by helpful ankle exoskeleton torques, and observed human adaptations to the altered cost landscape. These initial findings suggest emulators may indeed accelerate biomechatronic device development.

IV. Implications for Mechatronics

Much of what walking humans want from mechatronics remains a mystery, but some ideas can be inferred:

Technologies for Testbeds: Our testbeds show promise, but could be improved in terms of worn mass, peak torque, and especially closed-loop torque bandwidth. The more closely these tools mimic the autonomous devices they model, the more useful the experimental results. Laboratory settings allow for many actuation and control possibilities.

Mobile Device Hardware: Although product details must necessarily follow careful (emulated) experiments, some features are likely to be beneficial. Torque control is a necessity for fluid human-robot interactions. Low electrical energy consumption is needed for autonomy, and would be serviced by (torque controllable) energy recycling because most locomotor tasks are net energy neutral.

[1] A. D. Kuo, "Energetics of actively powered locomotion using the simplest walking model," *J. Biomech. Eng.*, vol. 124, pp. 113–120, 2002.

[2] J. E. A. Bertram and A. Ruina, "Multiple walking speed-frequency relations are predicted by constrained optimization," *J. Theor. Biol.*, vol. 209, pp. 445–453, 2001.

[3] S. H. Collins *et al.*, "Efficient bipedal robots based on passive-dynamic walkers," *Science*, vol. 307, pp. 1082–1085, 2005.

[4] S. H. Collins and A. Ruina, "A bipedal walking robot with efficient and human-like gait." in *Proc. Int. Conf. Rob. Autom.*, 2005, pp. 1983–1988.

[5] P. Bhounsule *et al.*, "A robot that can walk far using little energy," *Int. J. Rob. Res.*, 2012, submitted.

[6] S. H. Collins and A. D. Kuo, "Recycling energy to restore impaired ankle function during human walking," *PLoS: ONE*, vol. 5, p. e9307, 2010.

[7] J. M. Donelan *et al.*, "Biomechanical energy harvesting: Generating electricity during walking with minimal user effort," *Science*, vol. 319, pp. 807–810, 2008.

[8] H. M. Herr and A. M. Grabowski, "Bionic ankle-foot prosthesis normalizes walking gait for persons with leg amputation," *Proc. Roy. Soc. Lon. B*, vol. 279, pp. 457–464, 2012.

[9] iWalk, "The iwalk biom prosthetic foot and ankle," 2012, http://www.iwalkpro.com/.

[10] S. H. Collins, M. Wisse, and A. Ruina, "A three-dimensional passive-dynamic walking robot with two legs and knees," *Int. J. Rob. Res.*, vol. 20, pp. 607–615, 2001.

[11] S. H. Collins, P. G. Adamczyk, and A. D. Kuo, "Dynamic arm swinging in human walking," *Proc. Roy. Soc. Lon. B*, vol. 276, pp. 3679–3688, 2009.

[12] P. G. Adamczyk, S. H. Collins, and A. D. Kuo, "The advantages of a rolling foot in human walking," *J. Exp. Biol.*, vol. 209, pp. 3953–3962, 2006.

[13] M. Srinivasan, "Fifteen observations on the structure of energy minimizing gaits in many simple biped models," *J. Roy. Soc. Int.*, vol. 8, pp. 74–98, 2011.

[14] J. M. Wang *et al.*, "Optimizing locomotion controllers using biologically-based actuators and objectives," *Proc. SIGGRAPH*, vol. 31, 2012.

[15] J. Hidler *et al.*, "Multicenter randomized clinical trial evaluating the effectiveness of the Lokomat in subacute stroke," vol. 23, pp. 5–13, 2009.

[16] G. S. Sawicki and D. P. Ferris, "Mechanics and energetics of level walking with powered ankle exoskeletons," *J. Exp. Biol.*, vol. 211, pp. 1402–1413, 2008.

[17] J. F. Veneman *et al.*, "Design and evaluation of the LOPES exoskeleton robot for interactive gait rehabilitation," *Trans. Neural Syst. Rehabil. Eng.*, vol. 15, pp. 379–386, 2007.

[18] W. C. Flowers and R. W. Mann, "An electrohydraulic knee-torque controller for a prosthesis simulator," *J. Biomech. Eng.*, vol. 99, pp. 3–9, 1977.

[19] J. M. Caputo and S. H. Collins, "An experimental biomechatronic testbed for rapid assessment of robotic gait interventions," *J. Biomech. Eng.*, vol. submitted, 2012.

[20] J. L. Su and J. B. Dingwell, "Dynamic stability of passive dynamic walking on an irregular surface," *J. Biomech. Eng.*, vol. 129, pp. 802–810, 2007.

[21] M. Snaterse *et al.*, "Distinct fast and slow processes contribute to the selection of preferred step frequency during human walking," *J. Appl. Physiol.*, vol. 110, pp. 1682–1690, 2011.

[22] A. Roy *et al.*, "Robot-aided neurorehabilitation: a novel robot for ankle rehabilitation," *Trans. Rob.*, vol. 25, pp. 569–582, 2009.

[23] R. L. Burse and K. B. Pandolf, "Physical conditioning of sedentary young men with ankle weights during working hours," *Ergonomics*, vol. 22, pp. 69–78, 1979.

[24] K. E. Gordon, G. S. Sawicki, and D. P. Ferris, "Mechanical performance of artificial pneumatic muscles to power an ankle-foot orthosis," *J. Biomech.*, vol. 39, pp. 1832–1841, 2006.

[25] S. K. Au, J. Weber, and H. Herr, "Biomechanical design of a powered ankle-foot prosthesis," in *Proc. Int. Conf. Rehab. Rob.*, 2007, pp. 298–303.

[26] J. Hitt *et al.*, "Robotic transtibital prosthesis with biomechanical energy regeneration," *Indust. Rob: Int. J.*, vol. 36, pp. 441–447, 2009.

[27] M. Whittle, *Gait Analysis: An Introduction*. Oxford: Butterworth-Heinemann Medical, 1996.

[28] D. A. Winter, *Biomechanics and Motor Control of Human Movement*, 2nd ed. Toronto, Canada: John Wiley & Sons, Inc., 1990.

[29] G. C. Agarwal and G. L. Gottlieb, "Oscillation of the human ankle joint in response to applied sinusoidal torque on the foot," *J Physiol.*, vol. 268, pp. 151–176, 1977.

[30] J. E. Pratt *et al.*, "The roboknee: An exoskeleton for enhancing strength and endurance during walking," in *Proc. Int. Conf. Rob. Autom.* New Orleans, LA: IEEE, 2004, pp. 2430–2435.

[31] M. Noël *et al.*, "An electrohydraulic actuated ankle foot orthosis to generate force fields and to test proprioceptive reflexes during human walking," *Trans. Neural Syst. Rehabil. Eng.*, vol. 16, pp. 390–399, 2008.

[32] J. S. Sulzer *et al.*, "A highly backdrivable, lightweight knee actuator for investigating gait in stroke," *Trans. Rob.*, vol. 25, pp. 539–548, 2009.

[33] A. H. A. Stienen *et al.*, "Design of a rotational hydroelastic actuator for a powered exoskeleton for upper limb rehabilitation," *Trans. Biomed. Eng.*, vol. 57, pp. 728–735, 2010.