Supplementary Material

We used the Controlled Energy Storage and Return (CESR) prosthetic foot as a tool to explore the effects of altering ankle/foot mechanics, specifically Push-off work, on whole-body walking mechanics and metabolic cost. We were able to systematically vary energy storage and return in the foot by testing subjects walking on the CESR prosthesis with different energy-recycling springs (Table S1). Function of the CESR foot and experimental protocols for Amputee and Non-Amputee subjects are detailed below.

CESR Prosthetic Foot

The CESR foot is an energy recycling prosthesis capable of capturing energy elastically after heelstrike, storing the energy through mid-stance and returning the energy at terminal stance in the form of plantarflexion push-off work (Fig. 1). The energy recycling concept is based on dynamic walking principles [10], [11], [28] and human gait experiments [17] that suggest Push-off is the most economical time in the gait cycle to perform positive work because it minimizes mechanical Collision losses. Compared to conventional passive prosthetic feet, increased prosthetic ankle/foot Push-off work of the CESR foot more closely emulates the function of the natural ankle during walking. Energy recycling is accomplished by loading a compression die spring under the heel during early stance, locking the spring energy into place with a one-way clutch, then releasing a second latch that returns the spring energy to the forefoot for Push-off during the end of stance (Fig.1). The mechanism then resets during Swing phase. In addition to energy recycled from the CESR spring compression, Pre-load energy is also stored and returned in the forefoot, a cantilevered carbon fiber keel similar to those used in conventional dynamic elastic response prostheses. The prototype used in these studies was designed with separate toe and heel sections, each of which can articulate about a main mediolateral axis of the foot. A low-power microcontroller (worn on a small backpack with battery) used information from on-board angular displacement sensors at the heel and toe to determine activation timing of small motors that controlled energy storage and return. The mechanism and control design of the CESR foot are fully detailed in Collins and Kuo (2010) [12].

Amputee Protocol

Amputee subjects were recruited through the Seattle Veteran Affairs Hospital based on the following inclusion criteria: unilateral, transtibial amputation, age 18-80, weighing 70-100 kg, prosthesis user for minimum of 2 years, active and independent ambulator (i.e., no upper-limb aids), no history of injurious falls within the previous 6 months and free from neurological deficits or other musculoskeletal disorders. Seven subjects were initially recruited, but two were excluded because they were unable to complete the full protocol. This study was approved by the Veterans Affairs' Institutional Review Board and all subjects gave informed consent prior to participation.

Amputee subjects (N=5, age 50 ± 13 years, 76.7 ± 3.3 kg, leg length 0.97 ± 0.02 m) were tested at the Center of Excellence for Limb Loss Prevention and Prosthetic Engineering in Seattle, WA. Each subject performed acclimation and testing protocols on separate days, with the first day serving solely as an acclimation and training period. Treadmill acclimation was accomplished by 5 minutes of walking (1.14 m/s) on the subject's own prescribed foot. Next, about 5 minutes of overground and 5 minutes of treadmill walking were performed on the CESR foot for each of the three spring conditions and on the Conventional foot. The Conventional foot was worn and aligned inside a shoe, but since it was weight-matched to the CESR foot it was not intended to be a precise clinical baseline, only a qualitative control. All prosthetic alignment was performed by the same experienced prosthetist. In addition to the prosthesis, during all conditions subjects wore a small backpack (0.80 kg) containing a battery and microcontroller, which was connected to the prosthesis via ribbon cable and to the analog data acquisition via coaxial cable. To ensure full recovery after acclimation, at least 1 day separated training and collection sessions for all subjects. To ensure full retention of training, this recovery period was typically also fewer than 3 days, though in some cases was as high as 7 days due to Amputee subject availability.

During the data collection session, we recorded kinematic, kinetic and metabolic data for subjects walking on the CESR foot at 1.14 m/s at self-selected step frequency. Testing of the three spring conditions– Hard, Medium and Soft-PC – was randomized. For each condition we collected metabolic data, immediately followed by kinematic and kinetic measurements. Initially, a resting metabolic baseline was collected for 6 minutes during quiet standing. Next, treadmill metabolic testing lasted 10 minutes, the last 3 minutes of which were analyzed as steady state. Metabolic energy expenditure was approximated from oxygen and carbon dioxide exchange rates (i.e., indirect calorimetry; Brockway 1987) collected by the Oxycon Mobile wireless ergospirometry system (Viasys Healthcare, Yorba Linda, CA). In addition, a manual harness system was used during treadmill walking for safety reasons; however, the cable remained slack as not to interfere with gait. No adverse events that required engagement of the harness occurred for any subjects.

Kinematic and ground reaction force data were measured while subjects walked overground across force plates embedded along a 10 m walkway. Gait kinematics were collected with a 12-camera Vicon motion capture system (Oxford Metrics, Oxford, England) sampled at 120 Hz. We placed thirty-five 14 mm reflective markers on each subject at locations consistent with Vicon Plug-in-Gait full-body model (Oxford Metrics; Oxford, England). We placed an additional 4 markers on the CESR foot (heel, toe, medial/lateral articulating axis) to track motion of the foot segments. Ground reaction forces were collected with 2 Bertec force plates (Columbus, Ohio) and 2 AMTI force plates (Arlington, VA) sampled at 1200 Hz. A minimum of 6 successful overground trials were collected for each condition. We defined successful trials by the following two criteria: (1) walking speed was within range 1.14 ± 0.11 m/s as measured by photo gates and (2) at least two sequential foot strikes occurred on separate force plates.

Gait data were filtered (Woltring with mean-square-error value of 20), then standard 3D inverse dynamics were calculated using Vicon Plug-In-Gait dynamic model (Oxford Metrics; Oxford, England). A stride was defined from heelstrike to subsequent ipsilateral heelstrike based on gait events determined from Vicon's event detection algorithm. Inter-segmental power and COM work rate were calculated from forces and kinematics filtered at 25 Hz (Butterworth, 3rd order).

<u>*Non-Amputee Protocol*</u>

The Non-Amputee study was performed using similar methods to the Amputee study, but with some methodological differences due to different subject groups and equipment available at each testing site. The notable methodological differences are that in the Non-Amputee study:

(1) kinematics, kinetics and metabolic cost were recorded simultaneously because of availability of an instrumented force treadmill
(2) an additional control condition, shod walking in street shoes, was performed
(3) prosthetic simulator boot and lift shoe were worn by Non-Amputee subject
(4) subjects walked at slightly faster speed (1.25 vs. 1.14 m/s)
(5) the Medium and Soft-PC springs were identical, but the Hard spring was stiffer (324 vs. 262 N/mm due to comfort-related issues in the Amputee group)
(6) the conventional prosthetic foot (control condition) was worn without a shoe
(7) the subject group was considerably younger (24 ± 3 vs. 50 ± 13 years)
(8) subjects walked at fixed metronome frequency in all CESR spring condition trials

Non-Amputee subjects were recruited at the University of Michigan based on the following inclusion criteria: age 18-50, weighing 70-100 kg and with no known gait or balance impairments. The study was approved by University of Michigan Institutional Review Board and all subjects gave informed consent prior to participation.

All Non-Amputee subjects (N = 11, age 24 ± 3 years, mass 79.6 ± 7.1 kg nominally, 82.6 ± 7.1 kg while wearing prosthesis and accessories, leg length 0.97 ± 0.04 m nominally, leg length 1.08 +/- 0.04 m while wearing simulator boot) were tested in the Human Neuromechanics Laboratory at the University of Michigan. Subjects underwent training that was similar to the Amputee protocol. Training occurred 2 days prior to data collection. Subjects wore a prosthetic simulator boot (1.30 kg) with CESR foot (1.37 kg) attached unilaterally on the right leg and wore a lift shoe (1.42 kg) on the left leg to account for additional height (0.13 m) of the prosthesis. The simulator boots were modified AirCast© boots that immobilize the ankle and provide prosthetic attachment beneath the foot [17], [28]. Subjects wore a microcontroller and battery backpack identical to that described in the Amputee protocol. Spring conditions were changed without removing the prosthesis from the simulator boot in order to achieve consistent prosthetic alignment across conditions.

Kinematic, kinetic and metabolic data were measured simultaneously while subjects walked at 1.25 m/s on a custom-built instrumented treadmill. Conditions included baseline resting during quiet standing (6 minutes), walking in normal street shoes (10 minutes) and energy recycling spring condition trials (10 minutes). All walking conditions were randomized. Oxygen consumption and carbon dioxide production rates were measured using an open-circuit respirometry system (Physio-Dyne Instrument, Quogue, NY). Kinematic data were recorded at 120 Hz using an 8-camera motion capture system (Motion Analysis Corporation, Santa Rosa, CA). Markers were placed bilaterally according to a modified Helen Hayes marker set. Markers were placed on the simulator boot in locations approximating the anatomical bony landmarks and markers were rigidly attached to the prosthesis as described in the Amputee protocol. Ground reaction force data were recorded at 1200 Hz using a custom-built instrumented split-belt treadmill (Collins et. al., 2008). Kinematic and force data were filtered at 25 Hz. Standard 3D inverse dynamics were computed using custom software.

Data Analysis

We estimated mechanical power from measured kinematics and forces, and estimated metabolic energy consumption from oxygen consumption and carbon dioxide production. We made comparisons between spring conditions within a single group. Trends from the Amputee vs. Non-Amputee studies were qualitatively compared given the confounding methodological differences discussed above. We used the following mechanic and metabolic metrics:

1. *Center-of-mass work rate* was calculated independently for each limb. The work rate was computed from the 3D dot product of each limb's ground reaction force with COM velocity [29]. COM velocity was calculated from integration of ground reaction forces, assuming steady-state, periodic strides. This assumption was particularly strong for trials using an instrumented treadmill.

2. *Inter-segmental power* between foot and shank, i.e. power attributed to the prosthesis, was calculated based on summing translational and rotational work rates at the distal shank [30], [31], [35]. Translational power was calculated as the dot product of ankle joint force and translation velocity of the ankle marker (lateral malleolus). Rotational power was calculated as the dot product of the ankle moment (about the axis at the lateral malleolus) and angular velocity of the shank. This inter-segmental power method was used to estimate ankle/foot power in place of inverse dynamics because the former makes no rigid-body assumptions

about the prosthetic feet, whereas inverse dynamics is poorly suited to capture the unmodeled degrees of freedom in prosthetic feet (e.g., heel and toe keel deformations). In particular, standard inverse dynamic models fail to capture work performed about the independently articulating heel in the CESR prosthesis.

3. *CESR energy-recycling spring power (or rotational heel power)* was a supplementary metric used to approximate energy stored in the compression spring. This heel power was calculated from multiplying ankle moments with heel angular velocities, the latter derived from the on-board angular displacement sensor (a non-contact potentiometer).

4. *Joint power* about the ankle, knee and hip were computed from standard inverse dynamics as described previously in the Amputee and Non-Amputee protocols. Anthropometric data were modified to reflect changes in foot mass due to the CESR foot, simulator boot, and lift shoe, as appropriate for each subject and condition. Prosthetic ankle joint power from inverse dynamics was not reported (see #2 above on inter-segmental power)

5. *Metabolic power* was estimated from oxygen consumption and carbon dioxide production during steady-state gait, which we chose as the last 3 minutes of each walking trial. Power calculations were based on standard indirect calorimetry equations relating substrate metabolism to energy production [26]. Net metabolic results are presented, meaning that the resting metabolic rate was subtracted from metabolic power for each walking condition. As verification, we confirmed that respiratory exchange ratios were less than one for all subjects and conditions, indicating that energy was supplied primarily by aerobic metabolic pathways.

For each mechanical estimate, work summary measures were calculated by integrating under the power curves during specific phases of the gait cycle (Fig. S1). These phases or integration regions – Collision, Rebound, Pre-load, Push-off and Swing – were based on alternating regions of positive and negative COM work (Fig. 3). At the level of the prosthesis, energy-recycling spring energy storage was integrated over the negative power region in early stance, which in some cases was longer than the COM-defined Collision phase (Fig. 2). Energy return was defined as the integral over the positive power region preceding toe-off. All statistical comparisons were performed using a repeat measures ANOVA with Holm-Sidak correction. Nominally, P-values less than 0.05 were considered statistically significant, but with correction alpha = 0.017.

All mechanical measures were non-dimensionalized, averaged across subjects, and then re-dimensionalized for reporting purposes. Normalization constants were based on units of mass M, leg length L, and gravitational acceleration g. Average power and work normalization constants for Amputees were $Mg^{3/2}L^{1/2}$ = 2319 W and MgL = 729 J, and for Non-Amputee subjects were 2632 W and 874 J.

Table S1: CESR energy-recycling spring stiffness (N/mm)

	Amputees	Non-Amputees
Hard	262	324
Medium	157	157
Soft-PC*	42	42

* Soft-PC spring was also pre-compressed by 25 mm.

Table S2: Stride Time (sec)

	Amputees	Non-Amputees
Hard	1.28 (0.05)	1.30 (0.03)
Medium	1.28 (0.05)	1.30 (0.04)
Soft-PC	1.31 (0.02)	1.30 (0.04)
Conventional[†]	*1.26 (0.06)*	*1.19 (0.06)*
Shod[†]	*N/A*	*1.17 (0.06)*
P-values (alpha=0.017)	All > 0.17	All > 0.028

[†]*Italicized conditions are provided for reference, but were not tested statistically.*

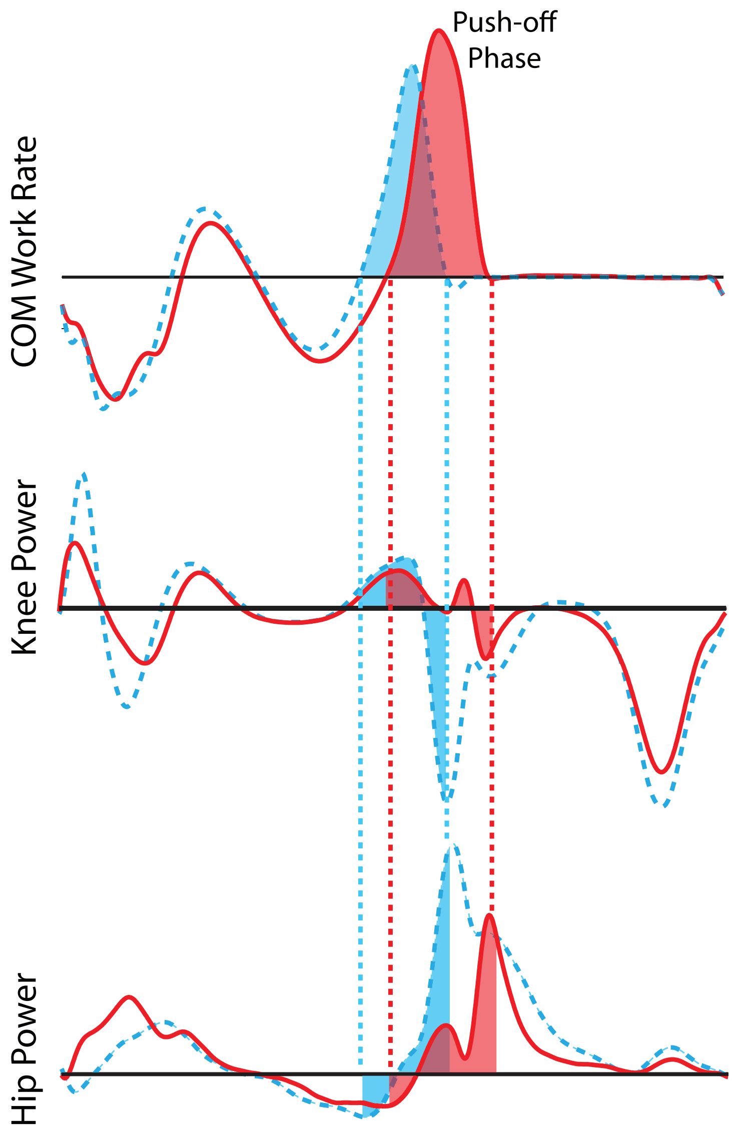

Figure S1: COM work rate and joint power over a stride: calculating work during Push-off. Phases of gait were defined independently for each subject and condition, based on alternating regions of positive and negative COM work rate. Knee and hip powers were integrated over shaded regions to compute joint work during Push-off phase. Plots are shown over a full stride from prosthetic limb heelstrike to heelstrike.

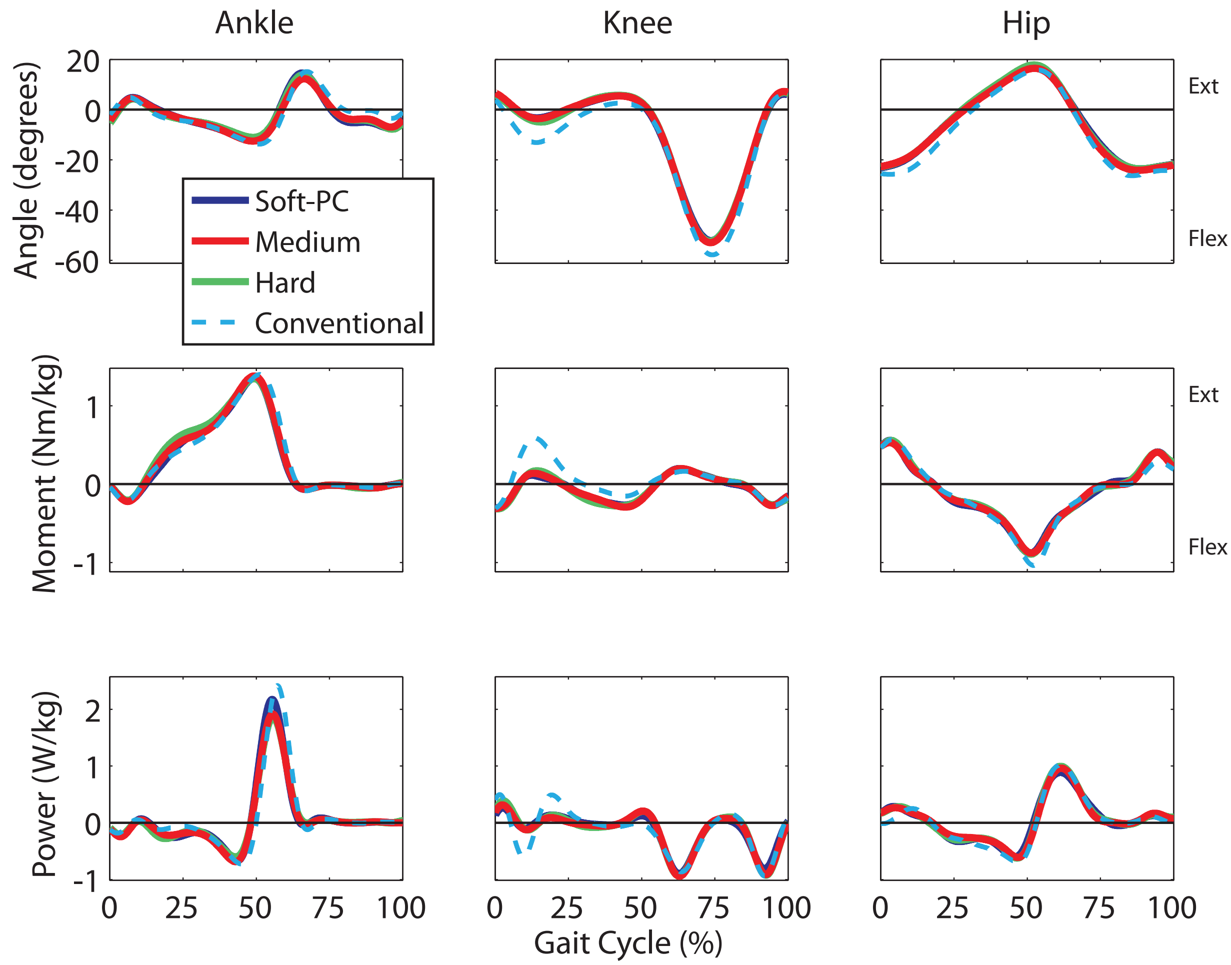

Figure S2: Average sagittal plane joint angles, moments and powers for ***Amputees' intact limb***. Plots are from intact limb heelstrike to heelstrike.

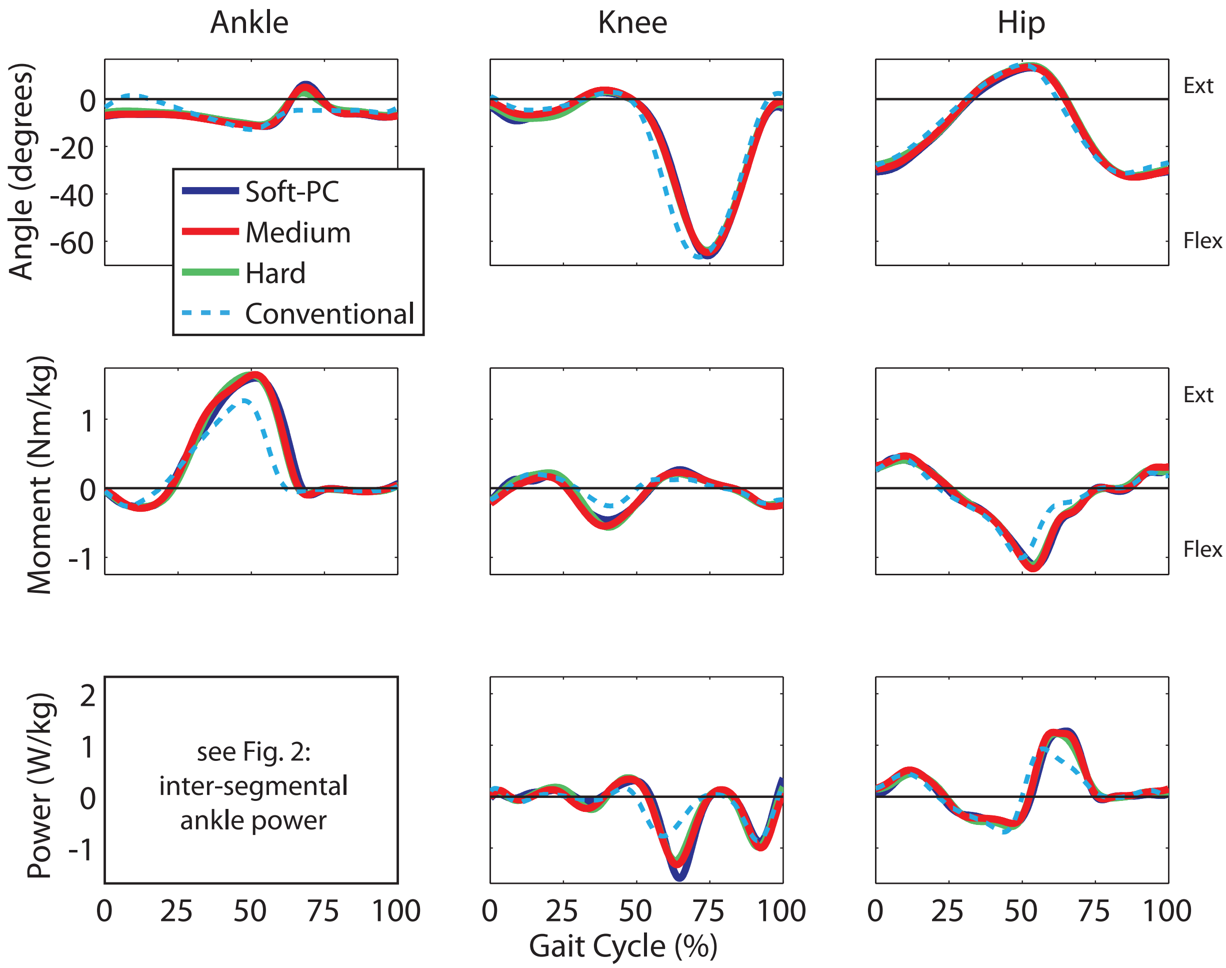

Figure S3: Average sagittal plane joint angles, moments and powers for ***Amputees' prosthetic limb***.

Plots are from prosthetic limb heelstrike to heelstrike.

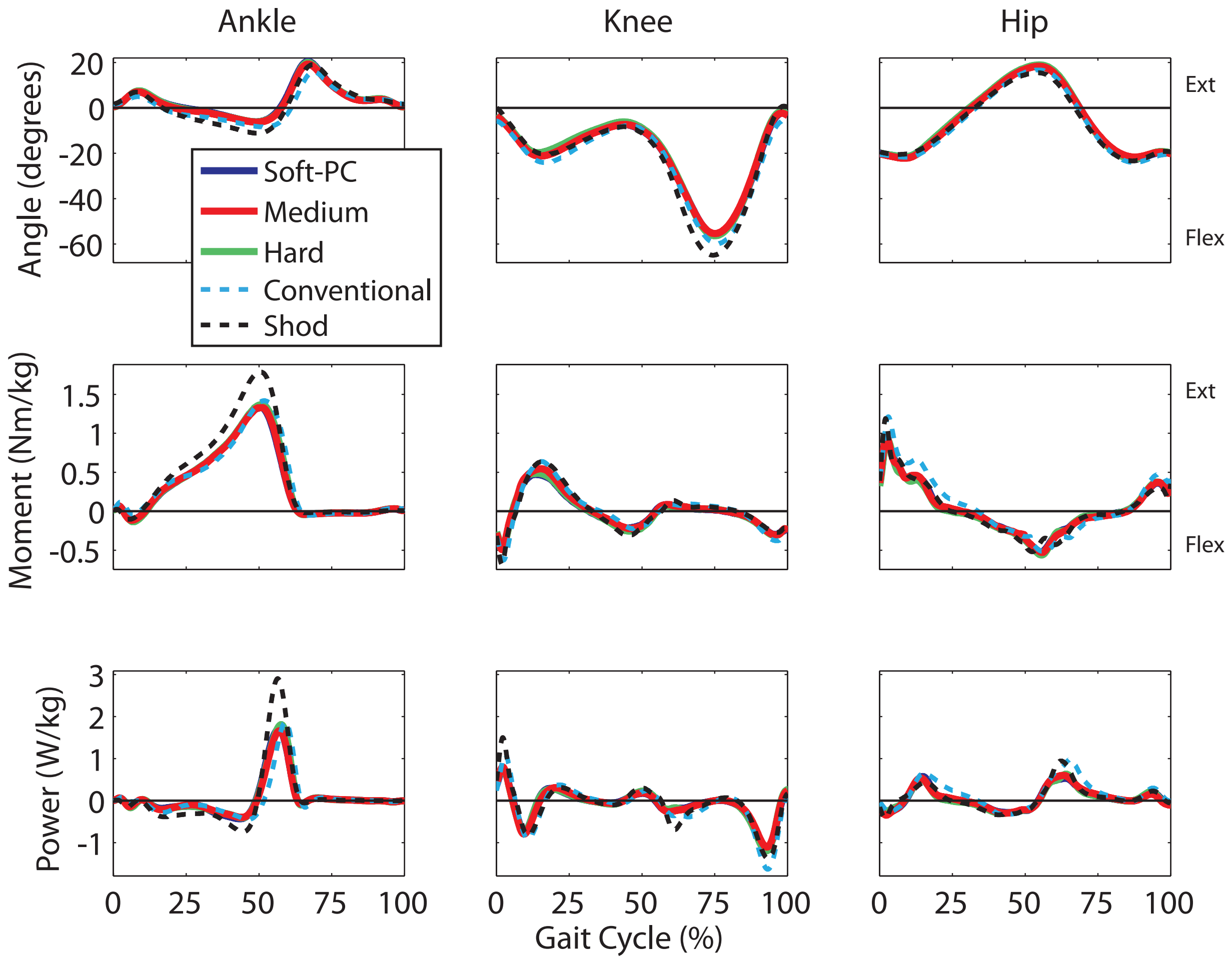

Figure S4: Average sagittal plane joint angles, moments and powers for ***Non-Amputees' intact limb***.

Plots are from intact limb heelstrike to heelstrike.

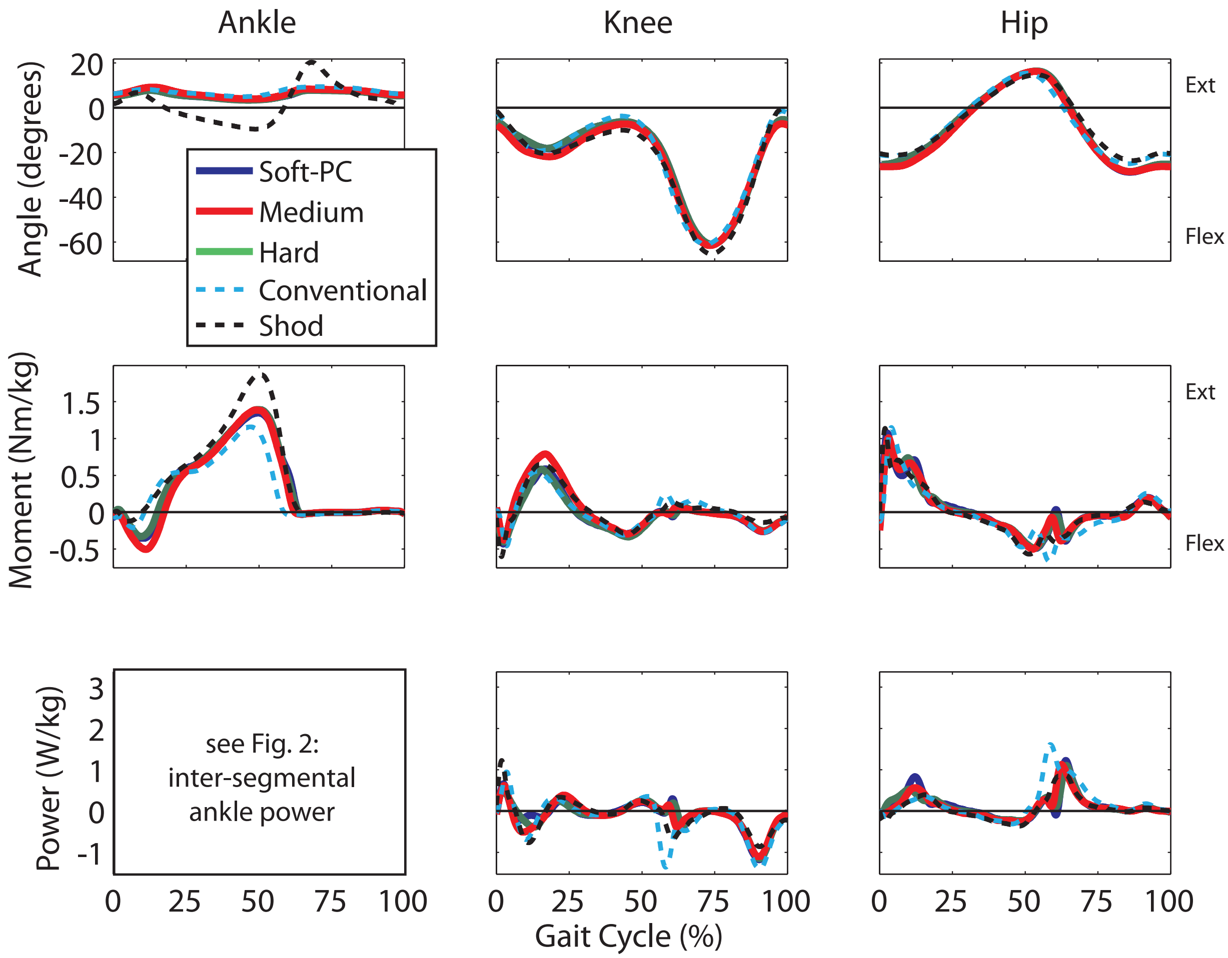

Figure S5: Average sagittal plane joint angles, moments and powers for ***Non-Amputees' prosthetic limb***. Plots are from prosthetic limb heelstrike to heelstrike.

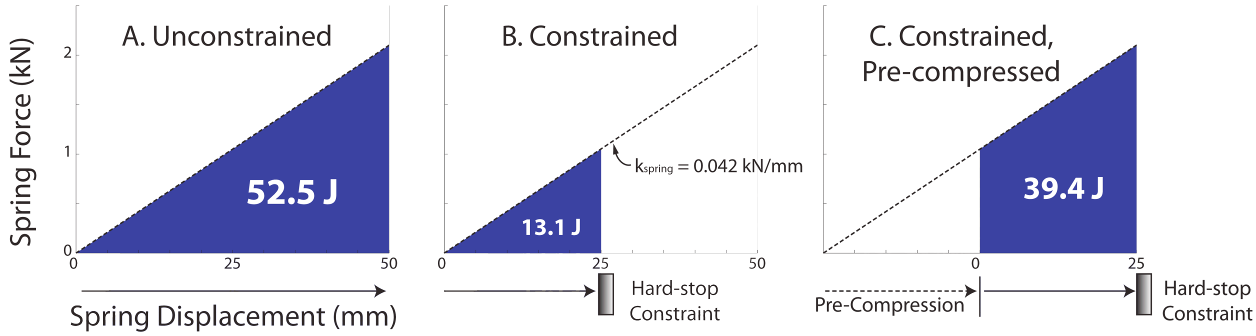

Figure S6: Effect of pre-compression on energy storage capacity of a spring. Example force-displacement curves are shown for a single spring under 3 different conditions: (A) unconstrained, (B) constrained and (C) pre-compressed and constrained. For (A) unconstrained springs (i.e., unlimited range over which they can compress), energy storage capacity is inversely proportional to spring stiffness (k_{spring}), so the softer the spring the more energy stored for a given load. An unconstrained spring of stiffness 0.042 kN/mm will deflect 50 mm when loaded with 2.1 kN, storing 52.5 J of energy. Practical limitations, however, such as the finite range of heel rotation in the CESR, often introduce a (B) hard-stop constraint. Assuming a maximum spring deflection constraint of 25 mm, the same spring would only be capable of storing 13.1 J (25% of the unconstrained spring capacity). By performing a (C) pre-compression on the spring, while keeping the maximum deflection constraint, the spring can store 39.4 J (75%). In summary, pre-compression allows a constrained spring to store more energy than it would otherwise be able to store.

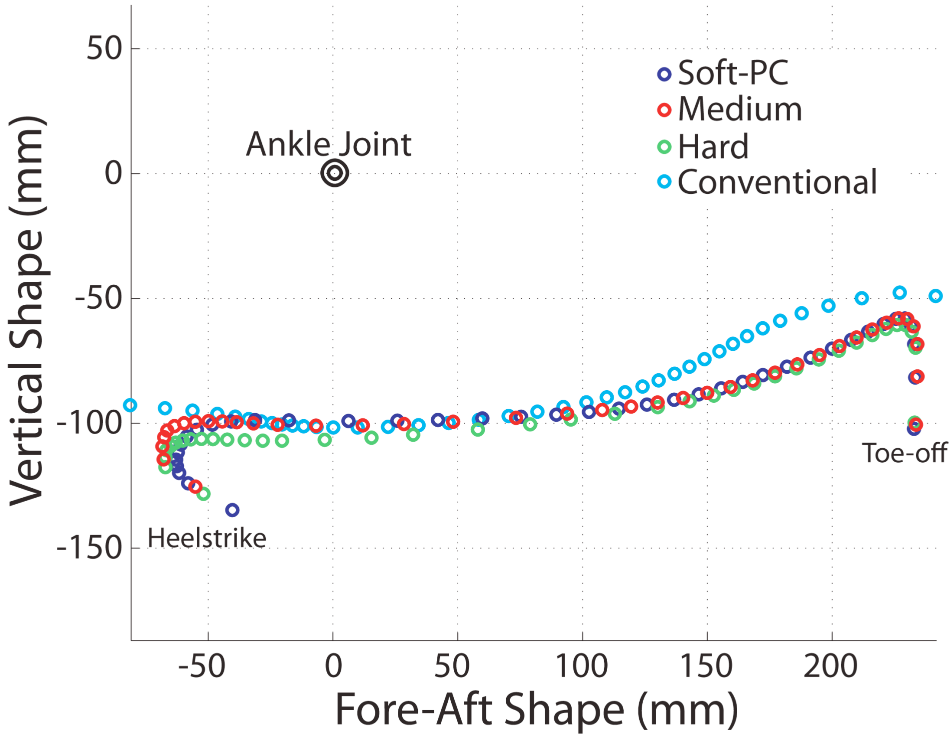

Figure S7: Rolling shape of the prosthesis. Average rolling foot shape (center of pressure plotted with respect to the ankle in the shank reference frame [26]) is presented for Amputee subjects. All CESR conditions appear similar during mid- to late-stance, but as expected some differences are observed during heel loading in early stance. We had no specific hypotheses regarding rolling shape. Implications of these differences are unclear and may require further study. Circles in plot represent samples at about every 0.02 seconds.