Is there a Trade-off between Economy and Task Goal Variability in Transfemoral Amputee Gait?

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Research

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Is there a trade-off between economy and task goal variability in transfemoral amputee gait?

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Abstract

Background: Energy cost minimization has been widely accepted to regulate gait. Optimization principles have been frequently used to explain how individuals adapt their gait pattern. However, there have been rare attempts to account for the role of variability in this optimization process. Motor redundancy can enable individuals to perform tasks reliably while achieving energy optimization. However, we do not know how the goal-irrelevant and goal-relevant variability is regulated.. In this study, we investigated how unilateral transfemoral amputees regulate step and stride variability based on the task to achieve energy economy.

Methods: Nine individuals with unilateral transfemoral amputation walked on a treadmill at speeds of 0.6,0.8,1.0,1.2 and 1.4 m/s using their prescribed passive prostheses. We calculated the step-to-step and stride-to-stride variability and applied goal equivalent manifold (GEM) based control to decompose goal-relevant and goal-irrelevant manifold. To quantify the energy economy, the energy recovery rate (R) was calculated based on potential energy and kinetic energy. Comparisons were made between GEM variabilities and commonly used standard deviation measurements. A linear regression model was used to investigate the trade-off between R and GEM variabilities.

Results: Our analysis shows greater variability along the goal-relevant manifold compared to the goal-irrelevant manifold ($p<.001$). Moreover, our analysis shows lower energy recovery rate for amputee gait compared to nonamputee gait (at least 20% less at faster walking speed). We found a negative relationship between energy recovery rate and goal-irrelevant variability. Compared to the standard deviation measurements, the variability decomposed using GEM reflected the preferred walking speed and the limitation of the passive prosthetic device.

Conclusion: Individuals with amputation cleverly leverage task redundancy, regulating step and stride variability to the goal equivalent manifold (GEM). This result suggests that task redundancy enables unilateral amputees to benefit from motor variability in terms of energy economy. The differences observed between prosthetic step and intact step support the development of prosthetic leg capable of enhancing positive work during the double support phase and of powered prosthesis controllers that allow for variability along the task space while minimizing variability that interferes with the task goal. This study provides a different perspective on amputee gait analysis and challenge the field to think differently about the role of variability.

Keywords: mechanical energy, goal equivalent manifold, transfemoral amputee, energy recovery rate, walking

Introduction

Energy cost minimization has been widely accepted to regulate gait. Experiments(1, 2) and computational models(3, 4) suggest that individuals select their preferred step length(4, 5), step width(6), step frequency(3), and walking speed(1, 2) to minimize the energy cost of walking. Energy optimization characterizes both the normal and the pathological gait, acting as an important factor in clinical applications(7, 8). Individuals with gait impairments operate close to their maximum level of effort and are prone to fatigue even at low walking speeds(9, 10). Hence, reducing the energy cost of walking is an important goal for rehabilitation(8, 11, 12) and a key metric in assessing the effectiveness of assistive devices such as prosthesis, orthosis, and exoskeletons (13, 14)

Optimization principles have been frequently used to explain how individuals adapt their gait pattern. However, there have been rare attempts to account for the role of variability in this optimization process. Variability is ubiquitous in motor performance (15, 16). Yet, a deviation from the average gait pattern might cause a divergence from energy optimality (17, 18). The fact that variation is inevitable raises a question that Is human variability regulated to assist the energy optimization process? If yes, then how?

Traditionally, movement variability has been linked to noise or error, and a large variability is considered “bad motor control”. This concept might be influenced by traditional statistical and assessment methods of movement variability (e.g., standard deviation) that assume randomness and independence of observations(16, 19). Opposite to this traditional perspective, variability can be seen to give rise to equifinality, allowing infinite number of ways to perform the same task(20, 21). Thus, individuals can tolerate movement variability that does not interfere with the task goal

without requiring extra effort (22, 23). These observations suggest that variability and movement redundancy can enable individuals to perform complex tasks reliably and repeatedly, increasing adaptability in motor performance (20, 24).

Researchers have analyzed movement redundancy and variability with different approaches, including uncontrolled manifold (UCM) analysis(24, 25), minimum intervention principle (MIP)(23, 26), and goal equivalent manifold (GEM)(21, 27). Despite methodological differences, all these approaches decompose the total variability at the body-level into goal-relevant variability and goal-irrelevant variability. The goal-relevant variability is the variability component tangent to the manifold, which does not affect the task goal. The goal-irrelevant variability is the variability component perpendicular to the manifold, which causes deviation from the task goal. Experiments focusing on quiet standing(24, 28), walking(29, 30), reaching(21, 23), and aiming(27) show that individuals preferentially constrains the goal-irrelevant variability rather than the goal-relevant variability, most likely because the latter is not detrimental to the achievement of the task goal. If we assume that goal-irrelevant variability interferes with the task goal and that this interference requires extra effort to correct, then higher goal-irrelevant variability should bel inked to increased energy cost. Based on this rationale, there may be a trade-off between energy optimization and regulation of task variability.

Understanding energy optimization requires us to determine how the features between the goal-irrelevant and goal-relevant variability emerge to reconcile with minimal energy cost. This problem is especially interesting in individuals with unilateral lower-limb amputations, who only have indirect control of the motion of the prostheses. Lower-limb amputees spend more metabolic energy than nonamputee individuals during walking and more proximal amputations are associated with greater metabolic energy cost than distal amputations(12, 31). The increased energy cost of

walking might be related to a lack of ankle push-off power from the prosthesis. Individuals with amputations compensate for the lack of push off power by pulling the thigh forward at the end of stance. This compensatory movement allows for continuous forward propulsion but is highly inefficient (14, 31, 32). In addition, walking with a passive prosthesis requires greater mechanical work for the step-to-step transition from prosthetic to intact limb, increasing the overall energy cost of walking.

Individuals with unilateral amputation need to adapt at every step to respond to external (i.e., lack of push-off) and internal (i.e., compensatory hip movements) perturbations. Thus, knowing how individuals control the step/ stride variability would be beneficial to understand how they achieve optimal energy cost. The purpose of this study is to investigate how individuals with unilateral transfemoral amputation walking with a passive prosthesis regulate step and stride variability. Specifically, we aim to understand how goal-relevant and goal-irrelevant variability are regulated with respect to energy economy. In this study, the amputee participants were asked to walk on a treadmill at different speeds using their prescribed passive prostheses. We applied GEM-based control to decompose the goal-relevant and goal-irrelevant variability(30, 33). Using this method, we found a goal-equivalent manifold containing all the possible stride/step time and stride/step length combinations that equally satisfy the goal function (details see Figure 1 and methods). To quantify the energy economy, we calculated the energy recovery rate for each step based on potential energy (E_p) and kinetic energy (E_k) (34). To optimize the recovery of mechanical energy, the E_p and E_k curves must have the same shape, be equal in amplitude, and be opposite in phase, as in a pendulum. A higher energy recovery rate indicates more efficient walking (see Figure 2B and methods for details).

We hypothesized that if individuals with amputations regulate their step/stride variability explicitly to minimize energy cost, then goal-relevant variability would be larger than goal-irrelevant variability given that these movement variations do not interfere with task performance. Moreover, we hypothesized that if individuals with amputations regulate their step/stride variability explicitly to minimize energy cost and the variability that interferes with task performance requires extra effort to correct, then there will be a trade-off between goal-irrelevant variability and the energy recovery rate across different walking speeds. To the best of our knowledge, this is the first study to analyze the gait variability at the level of task goal in connection to energy optimization in individuals with lower-limb amputations. Given that human behavior is goal driven, instead of focusing on common gait features such as step mean and standard deviation, we took the task goal into consideration. We believe that the results of this study will provide a different perspective on amputee gait analysis and challenge the field to think differently about the role of variability.

Methods

Experimental protocol and data collection

The data collection protocol has been carefully documented in Hood et al. (2020) (35), and data has been shared on Springer Nature, scientific data. The present study examined unilateral amputees walking on the treadmill who comfortably walked at or above 0.8 m/s were classified as full community ambulators (K3) (35). Seven male and two female transfemoral amputees (39.56±12.5 years old) participated after providing institutional-approved informed consent. The time since amputation was at least five years for all participants and participants self-reported using

the prosthesis for at least 5 hours a day. The details of the demographic data are provided in Table 1.

Each participant was asked to walk on a treadmill with their daily passive prostheses at speeds of 0.6, 0.8, 1.0,1.2 and 1.4 m/s. During acceleration and deceleration of the treadmill, participants were instructed to hold on to the treadmill handrails. After the treadmill reached the constant speed, the participants were encouraged to walk without holding the handrails and then recording started. For each treadmill speed, five trials contained around 10 continuous strides were recorded except for TF01 having four trials at 0.6, 0.8 and 1.0 m/s and TF02 having 4 trials at speed 0.6 m/s. Only TF5 and TF6 held the handrails all the times during the 1.4 m/s condition. During walking, a 10-camera motion capture system (Vicon, Oxford, UK) was used to record the reflective markers position at 200 Hz and a split-belt Bertec fully instrumented treadmill (Bertec Co; Columbus, OH) was used to record the bilateral ground reaction forces at 1000Hz. Reflective markers were placed on the participants following a modified Plug-in-Gait Model(36, 37). Marker trajectories and ground reaction force data were synchronized, recorded, and pre-processed using Vicon Nexus 2 software. A low-pass Butterworth filter with a cut-off frequency of 6 Hz was applied for the marker trajectories. Inverse dynamic of rotational, translational and potential energy of 15 segments (head, thorax, pelvis, and left and right forearms, upper arms, hands, thighs, shanks, feet) were calculated using Visual 3D.

Table 1 Inserts Around Here

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172 **Data analysis**

173 Stride-to-stride and step-to-step analysis were both performed in this study. For each
174 stride, n, stride time (T_n) was taken as the time between consecutive right heel strikes. Stride
175 length (L_n) was taken as the anterior–posterior distance travelled during each stride using the
176 position of right and left heel markers. Stride speed (S_n) was calculated as $S_n = L_n / T_n$. Step
177 analysis was calculated using the same methods and taken the time and length between heel
178 contact to the contralateral side heel contact. Prosthetic step was determined from prosthetic heel
179 contact to intact heel contact where the intact limb provided push-off power during the initial
180 double support phase and then transferred to the prosthetic single support phase. Intact step was
181 determined from intact heel contact to the prosthetic heel contact where the prosthesis leg
182 provided push-off power during the initial double support phase and then transferred to the intact
183 leg single support phase. (see Figure 2B).

184 **Goal Equivalent Manifold (GEM)**

185 We applied a GEM based method to decompose the goal-relevant and goal-irrelevant
186 variabilities on treadmill walking as originally proposed by Dingwell (33). The primary goal for
187 treadmill walking with speed *v* is to not walk off the treadmill. Thus, treadmill walking can be
188 formulated as:

189 $$-\frac{L_{TM}}{2} < \sum_{n=1}^{N} L_n - vT_n < +\frac{L_{TM}}{2} \qquad \text{(Equation 1)}$$

190 where n is each step/stride, L_n is step/stride length, T_n is step/stride time, the summation of $L_n -$
191 vT_n is the overall displacement walked over N steps and L_{TM} is the treadmill length. The simplest
192 strategy to satisfy Equation 1 is to keep v constant at each step, which was formulated using the

goal function:

$$L_n - vT_n = 0 \rightarrow \frac{L_n}{T_n} = v \qquad \text{(Equation 2)}$$

Hence, all $[L_n, T_n]$ pairs that satisfied Equation 2 defines the GEM, which was a red solid line in the L_n versus T_n plane (see Figure 1A).

To analyze walking dynamics relative to the GEM, we first normalized each L_n and T_n to unit variance ($\sigma = 1$) to provide an intuitive reference for comparison by dividing its own standard deviation:

$$\widetilde{T_n} = T_n / \sigma(T_n) \text{ and } \widetilde{L_n} = L_n / \sigma(L_n) \qquad \text{(Equation3)}$$

This yielded a GEM defined by the dimensionless walking speed: $\tilde{v} = mean(\widetilde{L_n} / \widetilde{T_n})$. We then define the new coordinates centered at a mean "preferred operating point":[$\widetilde{T_n}$ *, $\widetilde{L_n}$ *] = [mean($\widetilde{T_n}$), $\tilde{v}\widetilde{T_n}$*] and re-expressed as $\widetilde{T_n}' = \widetilde{T_n}$ - $\widetilde{T_n}$* and $\widetilde{L_n}' = \widetilde{L_n}$ - $\widetilde{L_n}$*.

Finally, the goal-equivalent deviation along the GEM (δ_T is the goal-relevant deviation), and non-goal-equivalent deviations perpendicular to the GEM (δ_p is the goal-irrelevant deviation) were calculated as:

$$\begin{bmatrix} \delta_T \\ \delta_p \end{bmatrix} = \frac{1}{\sqrt{1+v^2}} \begin{bmatrix} 1 & v \\ -v & 1 \end{bmatrix} \begin{bmatrix} \widetilde{T_n}' \\ \widetilde{L_n}' \end{bmatrix} \qquad \text{(Equation4)}$$

To quantify how variability was distributed relative to the GEM, the standard deviation of each new time series (δ_T and δ_p) was calculated (see Figure 1B).

Figure 1 Inserts Around Here

Energy recovery rate

The potential and kinetic energy of the whole body, divided unto *s* segments of mass *m*, can be measured from the gravitational and the kinetic energy of each segment calculated at each instant of time (*t*) relative to the frame of reference (Figure 2):

The whole-body potential energy was calculated as:

$$E_p(t) = \sum_{i=1}^{s}(m_i g h_i(t)) \qquad \text{(Equation 5)}$$

where h_i and m_i, respectively, are the vertical distance of center of mass to the ground and the mass of the ith segment, relative to the frame of global reference; g is the gravitational acceleration (9.81 m/s).

The whole-body kinetic energy was calculated as:

$$E_k(t) = \sum_{i=1}^{n}(\frac{1}{2}m_i\,[v_{xi}^2(t) + v_{yi}^2(t) + v_{zi}^2(t)] + \frac{1}{2}m_i\,[k_{xi}^2\omega_{xi}^2(t) + k_{yi}^2\omega_{yi}^2(t) + k_{zi}^2\omega_{zi}^2(t)])$$

(Equation 6)

where v_i, is the linear velocity relative to the frame of the global references in x, y and z axis; k_i and ω_i, respectively, are the moment of inertia about the axis through the center of mass and angular velocity of the ith segment (frame of local reference).

Thus, the total energy can be computed as:

$$\mathrm{E_{total}(t)} = \mathrm{E}_p(\mathrm{t}) + \mathrm{E}_k(\mathrm{t}) \qquad \text{(Equation 7)}$$

Over the stride and step period, the energy exchange was computed according to Winter(34). The energy recovery rate (R) represents the percentage of mechanical energy recovery via exchange between kinetic and potential energy in the COM movement. This is computed as:

$$R = 100 \cdot \frac{(w_{net} - w_{ext})}{w_{net}}$$

(Equation 8)

where $w_{ext} = \sum_{i=1}^{N} \left|\Delta E_p + \Delta E_k\right| \; during\ N\ sample\ period$, and $w_{net} = \sum_{i=1}^{N} \left(\left|\Delta E_p\right| + \left|\Delta E_k\right|\right)$ if one assumes no energy exchange between potential and kinetic energy to calculate the work done by a segment during N sample of period.

In an ideal energy recovery mechanism, the work associated with changes of potential energy is exactly the same as the work associated with kinetic energy changes, but with different sign: $W_p = -W_k$. That means that work produced to increase the potential energy can be obtained by reducing the kinetic energy and can again be returned to increase the kinetic energy at the next step-to-step transition. Hence, the larger value of R indicates better energy economy.

Figure 2 Inserts Around Here

Statistics

We first performed one-way repeated ANOVA to exam the commonly used gait features across walking speeds (mean and standard deviation (SD) of stride/ step length and time). The trend of the effect on speed was tested using within-subject contrast on a linear, quadratic, cubic, and order 4. To answer the question of the first hypothesis, the differences between the two types of decomposed variabilities (δT and δp) across five walking speeds were tested using 2-way repeated ANOVA (2 variabilities X 5 speeds) in stride, intact step, and prosthesis step. To answer the second question, we performed a simple linear regression of energy recovery rate (R) as the dependent variable and δp and walking speed as the independent variable for each stride, prosthetic step, and intact step (model: $R_i = \beta 0 + \beta 1\ \delta p_i + \beta 2\ speed_i + \varepsilon_i$). If the coefficient of δp is negative and reaches a significant level, it indicates a trade-off between R and δp. To understand the role of push-off for prosthetic and intact leg in respect to the energy exchange, a follow-up analysis was also conducted. We performed 2-way repeated ANOVA (2 (intact & prosthesis step) * 5 (Speeds)) to compare the energy recovery rate between step types and speeds. The significant level was set at α=.05. For the main effects that reached the significance level, the Bonferroni test was used for the post hoc comparisons.

Results

Primary gait features across walking speeds

Figures 3A and 3B depict the standard deviations (SD)of the length and time of prosthetic step, intact step, and stride, respectively. For the length SD of prosthetic step, intact step, and stride linearly decreased with faster walking speed (Figure3C) (prosthetic step: $F(4,32) = 11.629$, $p<.001$, $\eta_p^2 = .592$, linear effect $F(1,8) = 24.876$, $p<.001$, $\eta_p^2 = .757$; intact step: $F(4,32) = 6.826$, $p<.001$, $\eta_p^2 = .460$, linear effect $F(1,8) = 22.238$, $p<.001$, $\eta_p^2 = .735$; stride: $F(4,32) = 6.646$, $p<.001$, $\eta_p^2 = .454$, linear effect $F(1,8) = 14.889$, $p<.01$, $\eta_p^2 = .650$). The time variability of prosthetic step, intact step, and stride also linearly decreased with faster walking speed (Figure3D) (prosthetic step: $F(4,32) = 51.593$, $p<.001$, $\eta_p^2 = .866$, linear effect $F(1,8) = 158.259$, $p<.001$, $\eta_p^2 = .9527$; intact step: $F(4,32) = 29.234$, $p<.001$, $\eta_p^2 = .785$, linear effect $F(1,8) = 51.545$, $p<.001$, $\eta_p^2 = .866$; stride: $F(4,32) = 53.982$, $p<.001$, $\eta_p^2 = .871$, linear effect $F(1,8) = 117.232$, $p<.01$, $\eta_p^2 = .936$).

Figure 3 Inserts Around Here

Gait Variability Decomposed Based on GEM

The first hypothesis was whether the amputees exhibited greater variability along the GEM (task goal) rather than perpendicular to it. Figure 4A, 4B and 4C respectively demonstrate that the task relevant (δ_T) variability was significantly larger than task irrelevant (δ_P) variability in stride,

prosthetic step, and intact step (stride: $F(1,8) = 47.885$, $p<.001$, $\eta_p^2 = .857$; prosthetic step: $F(1,8) = 158.198$, $p<.001$, $\eta_p^2 = .952$; intact step: $F(1,8) = 172.667$, $p<.001$, $\eta_p^2 = .956$). For the interaction effect between speed and variability, there was a significant difference in prosthetic step ($F(4,32) = 6.827$, $p<.001$, $\eta_p^2 = .46$) and a marginal level on stride ($F(4,32) = 2.670$, $p=.051$, $\eta_p^2 = .26$). The post hoc analyses for δ_T in prosthetic step showed that at the speed of 1.0 m/s, 1.2 m/s and 1.4 m/s were equal but were significantly higher than the speed at 0.6 and 0.8 m/s. δ_p in prosthetic step, on the other hand, showed a trend as an U shape that walking speed at 1.0 m/s was significantly smaller than 0.6m/s and 0.8 m/s and significantly smaller than 1.2 m/s and 1.4 m/s. The post hoc analyses for δ_T in stride showed that 1.2 m/s was significantly smaller than 1.0 m/s and 1.4 m/s , and δ_p showed 1.2 m/s was significantly larger than 1.0 m/s and 1.4 m/s.

Figure 4 Inserts Around Here

Energy recovery rate

Figures 5 depicts the energy recovery rate of prosthetic step and intact step. There was a significant interaction between step types and speed ($F(4,32) = 5.552$, $p=.002$, $\eta_p^2 = .40$), and the post hoc analysis indicated only in prosthetic step at speeds 1.2m/s and 1.4 m/s were significantly larger than other speeds. The second hypothesis was whether there is a trade-off between δp and energy recovery rate (negative relationship). We performed a linear regression between δp, speed, and energy recovery rate of prosthetic step and intact step. The results showed the coefficient in

prosthetic step and intact step reached significant differences with a negative coefficient (See Table 2).

Figure 5 Inserts Around Here

Table 2 Inserts Around Here

Discussion

The purpose of this study was to investigate how individuals with unilateral transfemoral amputation regulate step and stride variability on the goal-relevant and goal-irrelevant manifold to achieve energy economy. We examined the hypothesis that if amputees are regulating variability in terms of task goal, a large portion of goal-relevant variability would be observed. In addition, we examined whether there would be a trade-off between goal-irrelevant variability and energy recovery rate across different walking speeds under the assumption that individuals require more energy to correct for the goal-irrelevant variability.

Looking at the main results of primary gait features, the standard deviation of stride/step length and time linearly decreased with faster walking speeds (Fig. 3A and 3B). This trend was consistent for both prosthesis steps and intact steps. While the standard deviation quantified the magnitude of deviation from the mean of all strides/steps, the time series of goal-relevant (δ_T) and goal-irrelevant (δ_p) variabilities exhibited the step-to -step change in respect to the task goal with the temporal order. δ_T, qualitatively, showed larger amplitudes than the δ_p (Figure 1B) and the variabilities of δ_T were significantly higher than the δ_p across all speeds (Figure 4). These results support our first hypothesis and indicate that amputees explicitly regulate step/stride variability tuned to the goal equivalent manifold (GEM). Our findings on unilateral amputees followed the same trend as healthy young adults(33, 38) and elderly(39, 40). The ability to leverage the task redundancy could be critical when tasks become more demanding. For example, walking on uneven terrain requires individuals to adjust each step rather change their average gait pattern. In addition, this concept can be used as a guiding principle for designing fault-tolerant controller for powered prostheses. To achieve this goal, a powered prosthesis controller should maximize the

343 variability along with the δ_T while minimizing the δ_p. Moreover, the powered prosthesis controller
344 should only correct errors distributed on the δ_p because these errors interfere with task performance.
345 In contrast, errors that vary along with the δ_T should be tolerated as they do not interfere with the
346 task performance A prosthesis controller following this principle may improve safety while
347 reducing control effort.

348 The goal-relevant (δ_T) and goal-irrelevant (δ_p) variabilities demonstrated a different speed
349 effect compared to the standard deviation of stride/step length and time, and such effects reflect
350 the unique characteristics of amputee gait (Figure 4). During the prosthetic step, the intact leg
351 regulates the push-off power, which propels the body forward, and the swing movement. Thus, the
352 intact leg largely determines the step time and step length. Our results show that the variability of
353 δ_T and δ_p changed with walking speed forming an inverted U and U shape, respectively (Figure
354 4B). A similar trend was observed with healthy adults(33, 38). This trend suggests that the intact
355 leg can adjust the δ_T and δ_p variabilities that tuned to the preferred walking speed for unilateral
356 amputees (around 1 m/s) (41). Interestingly, the intact step did not show the same adaptability to
357 walking speed. During the intact step, the passive prosthetic leg provides limited push-off power
358 (32, 42) and swing time adaptability. Thus, although microprocessor-controlled knee prostheses
359 can adjust the joint damping to walking speed(43, 44), our results indicate that this adaptation
360 mechanism is not enough to achieve similar task performance to the intact leg. Future work could
361 test whether the U-shaped relationship observed in the intact leg occur in powered prostheses
362 capable of imitating the intact-leg push-off powered and swing time adaptability.

363 In this study, we quantified energy recovery rate at each step based on potential energy (E_p)
364 and kinetic energy (E_k) (34). Previous studies has assessed the effectiveness of the pendulum

365 mechanism during human walking by measuring the fraction of the total mechanical energy that
366 is recovered as a result of the transduction between E_p and E_k(45, 46). To optimize the recovery
367 of mechanical energy, which leads to the most efficiency walking pattern, the E_p and E_k curves
368 must have the same shape, be equal in amplitude, and be opposite in phase, as in a pendulum. A
369 higher energy recovery rate indicates more efficient walking (see Figure 2B and methods for
370 details). Previous studies showed that for healthy adults at preferred walking speed (~ 1.6 m/s), as
371 much as 70% of the required external mechanical energy can be recovered due to this energy
372 saving mechanism (45, 47, 48). The other 30% of external mechanical energy is lost from the
373 system and must be supplied by the muscles(49). Our study shows that unilateral amputees have
374 lower recovery rate than healthy adults. At 1.4 m/s, the average energy recovery rate during the
375 prosthetic step was about 40%. Moreover, the recovery rate could be up to 60% depending on the
376 amputee individual (see Figure 5). This result explains why individuals with amputation using a
377 passive prosthesis spend, on average, more energy during walking than nonamputee individuals.

378 As expected, the prosthetic step had a larger energy recovery rate than the intact step.
379 Interestingly, the speed trends for the prosthetic and intact steps were different. The prosthetic step
380 showed a linearly increase of the energy recovery rate with walking speed, whereas the intact step
381 showed an inverted U-shaped profile (Figure 5). This difference in the speed trends could be due
382 to the different ability of the intact and prosthetic leg to generate positive work during push-off.
383 Previous studies in healthy adults found an inverted U-shaped relationship, although for a wider
384 range of walking speeds than the one used in this study (i.e., 0.5-2.5 m/s vs. 0.8-1.4 m/s) (45, 47,
385 48). In these studies, energy recovery rate gradually increased from 0.6 m/s to 1.6 m/s, which is
386 similar to the walking speed range of our study This result is in agreement with our finding in the
387 prosthetic side.

388 Previous studies in nonamputee individuals show that, at low walking speeds, the potential
389 energy changes are larger than the kinetic energy changes. When walking speed increases, the
390 kinetic energy changes increase relative to the potential energy changes, exceeding the potential
391 energy changes after the walking speed is above the preferred walking speed. Thus, the energy
392 recovery rate reaches its maximum for the self-selected speed and it is lower than its maximum
393 both above and below the self-selected speed (45). Our study shows that when the intact leg is in
394 charge of forward propulsion (i.e., prosthesis step), the amplitude of kinetic energy continuously
395 increases along with walking speed. However, when the prosthetic leg is in charge of forward
396 propulsion (i.e., intact-leg step), the capacity of increasing the amplitude of kinetic energy relative
397 to the potential energy reaches a limit after the walking speed is above the amputee's preferred
398 walking speed. We speculate that this inability to increase the amplitude of kinetic energy might
399 be due to the passive prosthesis generating less power than the intact leg during push-off, resulting
400 in net negative work during the intact step(32, 42, 50). The increase of energy dissipation during
401 double support phase at fast speed might explains the inverted U-shaped relationship in a smaller
402 range of walking speed. We speculate that, with proper control of propulsive torque, powered
403 prostheses(14, 42) could enable amputees to achieve better energy recovery rate. Future studies
404 should investigate the step-to-step change of energy recovery rate with powered prostheses to test
405 this hypothesis.

406 The second hypothesis was tested by applying linear regression to the time series of δ_p and
407 R calculated extracted from each stride and step. Using this regression, we found that adding the
408 δ_p in the linear model significantly explains the energy recovery rate. Moreover, the negative
409 coefficient of the linear regression supports that there is the trade-off relationship between δ_p
410 variability and recovery rate. Since the fitting considered the temporal order with δ_p and the energy

411 recovery rate, this result demonstrates that individuals with unilateral amputations leverage
412 movement redundancy to correct small deviations at each step to control the task variability while
413 minimizing energy cost. This short-term control mechanism (step-to-step) exploits the inherent
414 task redundancy enabling unilateral amputees to receive potential benefits from motor variability
415 in achieving energy economy. This result could inform the future design and control of powered
416 lower-limb prostheses. For example, a powered prosthesis controller that enables variations along
417 the goal-equivalent manifold may improve energy efficiency in amputee locomotion. This idea
418 could be the object of future work for rehabilitation engineers.

419 We recognize that the energy recovery rate might not accurately estimate the energy cost
420 of walking, especially for fast walking speeds (32). Although measuring the oxygen consumption
421 can lead to a more accurate estimate of the energy cost of walking, it cannot provide step-to-step
422 cost, which is needed for the proposed analysis of GEM variabilities. Using more elaborate
423 musculoskeletal models might provide a better estimation of the energy cost of walking; however,
424 inverse dynamics calculations have been shown not to solve this problem easily(51) and the current
425 validity of more elaborate musculoskeletal models does not warrant success with such an approach
426 either. We believe the interaction of task variability and energy recovery rate could provide some
427 clarification on the issue. Further interpretation would be highly speculative and require more
428 studies.

429 **Conclusion**

430 This study investigated how individuals with unilateral transfemoral amputation walking
431 with a passive prosthesis regulate step and stride variability on the goal-relevant and goal-
432 irrelevant manifolds. Our results suggest that individuals with amputation cleverly leverage task

redundancy, regulating step and stride variability to the goal equivalent manifold (GEM). This result suggests that task redundancy enables unilateral amputees to benefit from motor variability in terms of energy economy. The differences observed between prosthetic step and intact step support the development of prosthetic leg capable of enhancing positive work during the double support phase. Moreover, the results of this study motivate the development of powered prosthesis controllers that allow for variability along the task space while minimizing variability that interferes with the task goal. This study provides a different perspective on amputee gait analysis and challenge the field to think differently about the role of variability.

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Ethics approval and consent to participate

The experimental protocol for this study was approved by the University of Utah Institutional Review Board.

Consent for publication

All subjects provided written informed consent, including written permission to publish photos and videos of the experiment.

Availability of data and materials

The data has published at scientific data

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Contributions

IL and BLF processed the data. TL directed the data collection. ML and IL discussed the results. IL drafted the manuscript. HH, ML,BLF. and TL revised the manuscript. All authors read and approved the final manuscript prior to publication.

Competing interests

The authors declare no competing interests.

Figures

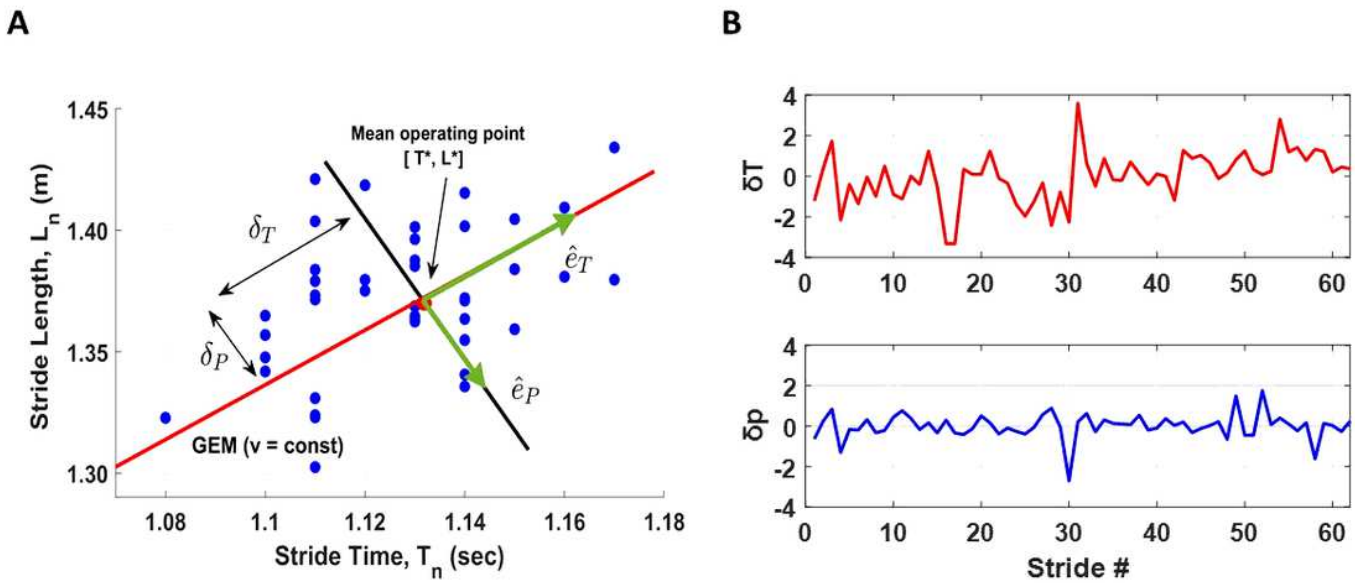

Figure 1

(A) Schematic depicting the goal equivalent manifold (GEM) for maintaining constant walking speed (v) using data from TF01. The red dot is the preferred mean operating point. Each blue dot represents the combination of stride time and stride length for an amputee participant with walking speed at 1.2 m/s. Dots that lie exactly on the red diagonal line (GEM) achieved the same speed and satisfied the goal. (B) The time series of δT and δp deviations for the data set shown in Figure 1A.

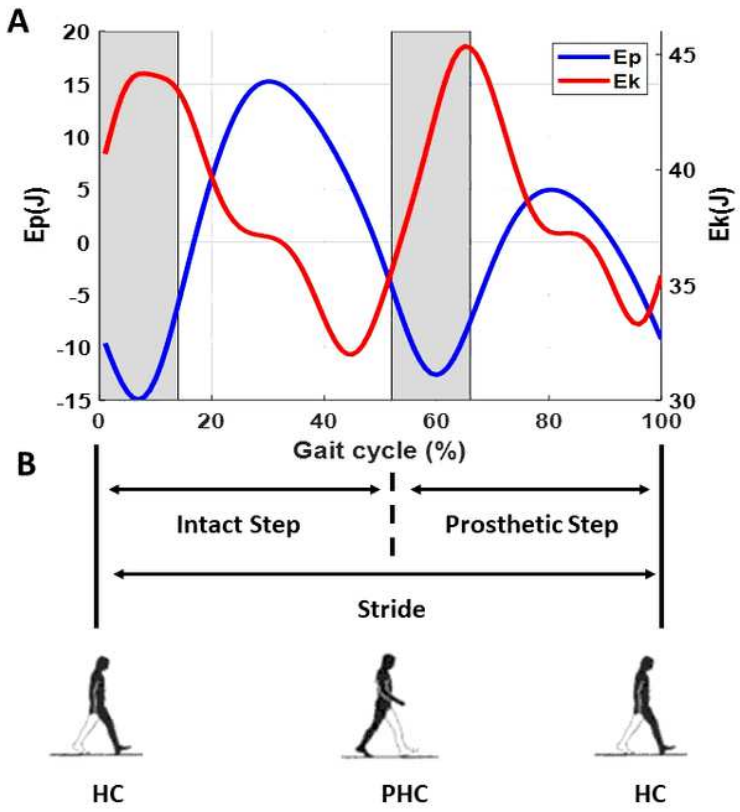

Figure 2

.(A) Example of Ep and Ek oscillations are plotted as a function of gait cycle duration. The shaded area indicates the periods of double support. The plot shows an out-of-phase oscillation of kinetic and potential energy, allowing energy exchange to occur between Ek and Ep . (B) Demonstration the stride and step (prothesis/intact step) period determined in this study. HC is heel contact of intact limb; PHC is heel contact of prosthetic limb. Stride period is determined from HC to the next HC. Prosthetic step is determined from PHC to HC where prosthetic limb leads, intact limb trails, and single support is on the prosthetic limb. Intact step is determined from HC to the PHC where intact limb leads, prosthetic limb trails, and single support is on the intact limb.

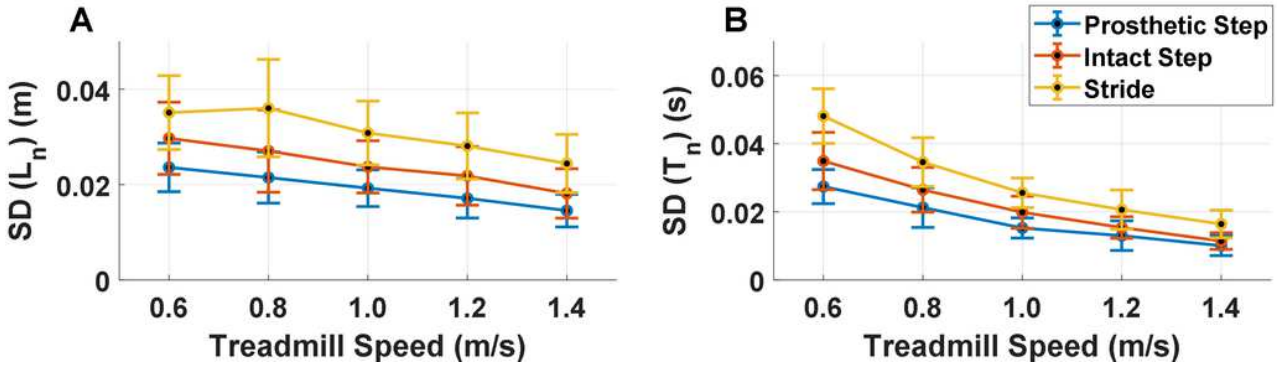

Figure 3

Primary gait parameters. Standard deviations (A, B) for stride and step length (Ln) and time (Tn) as a function of walking speed from 0.6 m/s to 1.4 m/s. Error bars indicate between subject 95% confidence intervals at each speed. The SD of length and the SD of time in stride and step length linearly decreased along with speed.

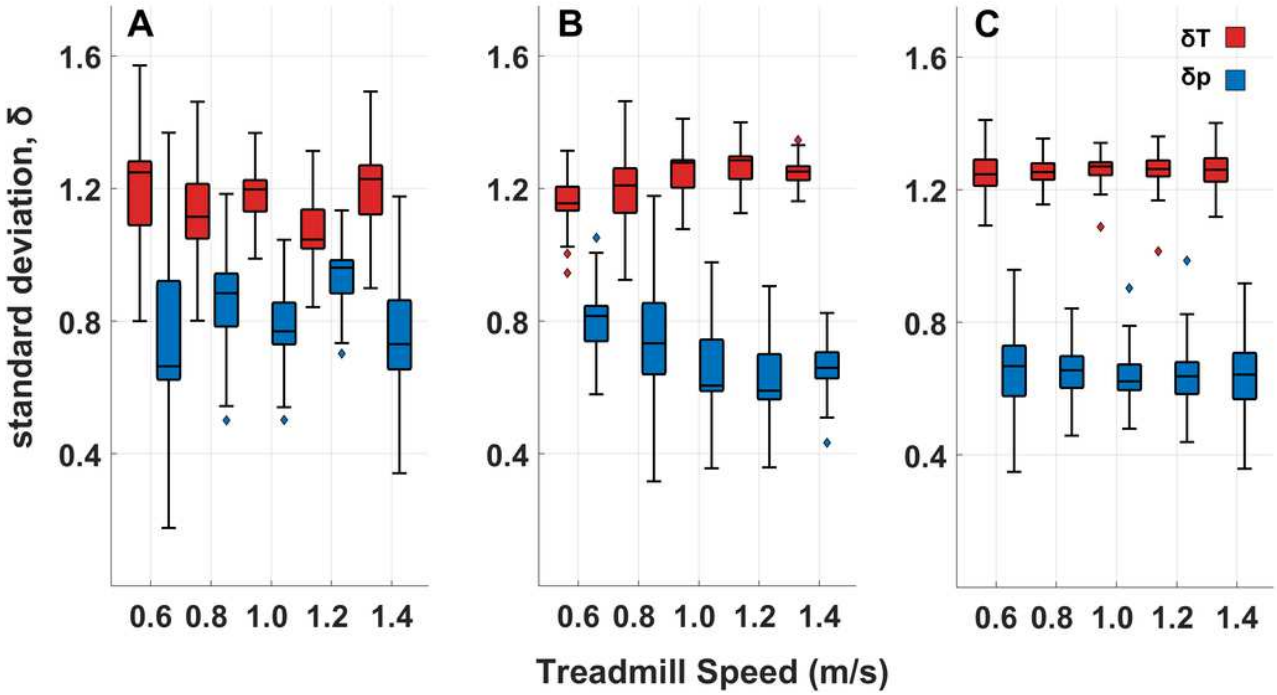

Figure 4

GEM-Based Decomposition of Gait Variability. (A). Standard deviations for all δT and δp time series at all 5 walking speeds calculated from each stride period. (B). Standard deviations for all δT and δp time series at all 5 walking speeds calculated from each prosthetic step. (C). Standard deviations for all δT and δp time series at all 5 walking speeds calculated from each intact step.

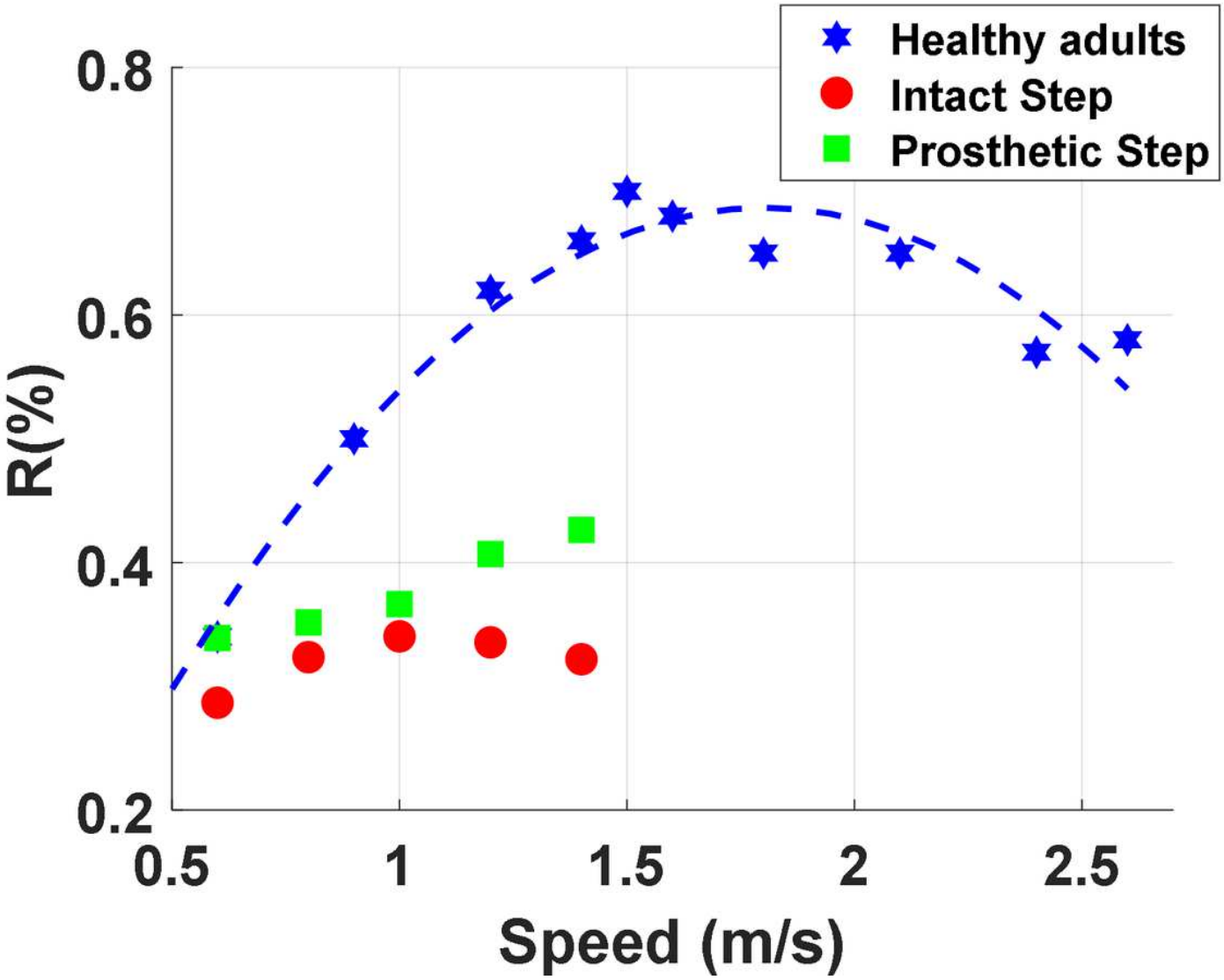

Figure 5

Energy recovery rate (R) are plotted as a function of walking speed for healthy adults (hexagram), Intact Step (circles) and Prosthetic Step (square). Adult data were reproduced from Willems et al., 1995.