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Dynamic stability of individuals with transtibial amputation walking in destabilizing environments ☆

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ABSTRACT

Lower limb amputation substantially disrupts motor and proprioceptive function. People with lower limb amputation experience considerable impairments in walking ability, including increased fall risk. Understanding the biomechanical aspects of the gait of these patients is crucial in improving their gait function and their quality of life. In the present study, 9 persons with unilateral transtibial amputation and 13 able-bodied controls walked on a large treadmill in a Computer Assisted Rehabilitation Environment (CAREN). While walking, subjects were either not perturbed, or were perturbed either by continuous mediolateral platform movements or by continuous mediolateral movements of the visual scene. Means and standard deviations of both step lengths and step widths increased significantly during both perturbation conditions (all $p < 0.001$) for both groups. Measures of variability, local and orbital dynamic stability of trunk movements likewise exhibited large and highly significant increases during both perturbation conditions (all $p < 0.001$) for both groups. Patients with amputation exhibited greater step width variability ($p = 0.01$) and greater trunk movement variability ($p = 0.04$) during platform perturbations, but did not exhibit greater local or orbital instability than healthy controls for either perturbation conditions. Our findings suggest that, in the absence of other co-morbidities, patients with unilateral transtibial amputation appear to retain sufficient sensory and motor function to maintain overall upper body stability during walking, even when substantially challenged. Additionally, these patients did not appear to rely more heavily on visual feedback to maintain trunk stability during these walking tasks.

1. Introduction

People with lower limb amputation often experience considerable difficulty walking (Grumillier et al., 2008) and decreased physical capacity (van Velzen et al., 2006). More than 50% of community-dwelling amputees fall each year (Miller and Deathe, 2004) and younger patients fall as often as older patients (Miller et al., 2001). Amputation disrupts motor and proprioceptive function, leading persons with amputation to rely more heavily on visual feedback for standing balance (Fernie et al., 1978; Vanicek et al., 2009). Visual and somatosensory feedback are also thought to be critical for walking (Gandevia, 2001). However, walking and standing are very different tasks (Kang and Dingwell, 2006) and passive dynamics likely play a greater role during walking (Srinivasan and Ruina, 2006; Iida and Tedrake, 2010). Patients with unilateral transtibial amputation retain intact visual and vestibular input, and mostly intact proprioception and motor function in their proximal impaired limb and contralateral limb. Thus, it remains an important question whether these patients retain sufficient sensory and motor capacity to maintain adequate dynamic stability while walking.

Here, we define dynamic stability as the capacity of a system to respond to perturbations (Strogatz, 1994). "Local stability" refers to responses to very small (i.e., "local" in state space) perturbations that may arise from either external (e.g., irregular surfaces) or internal (e.g., neuromuscular noise) sources. For aperiodic systems, this local stability is quantified by local divergence exponents (Dingwell and Cusumano, 2000). For periodic (i.e., limit cycle) systems, maximum Floquet multipliers quantify the local stability of consecutive orbits around the limit cycle, often termed "orbital stability" (Strogatz, 1994; Dingwell and Kang, 2007). Because

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human walking is neither strictly periodic, nor strongly aperiodic, quantifying both measures provides a more complete description of the sensitivity of the system to small perturbations.

Healthy elderly are more locally unstable than young adults (Kang and Dingwell, 2008a; Hamacher et al., 2011) and elderly fallers may exhibit both increased orbital (Granata and Lockhart, 2008; Hamacher et al., 2011) and local (Toebes et al., 2012) instability compared to non-fallers. Compared to nondisabled controls, greater local instability has been demonstrated in patients with either transfemoral (Lamoth et al., 2010) or transtibial (Hak et al., 2013; Wurdeman et al., 2013) amputation. However, these studies did not control for walking speed, which can independently alter local stability (Dingwell and Marin, 2006; England and Granata, 2007; Kang and Dingwell, 2008a; Manor et al., 2008). Slower walking speeds are commonly adopted by elderly (Tsai and Lin, 2013) and many patient populations, including those with amputation (Paysant et al., 2006; Kendell et al., 2010). Conversely, faster walking speeds generally increase fall risk (Pavol et al., 1999; Troy et al., 2008; Faulkner et al., 2009; Callisaya et al., 2012). Separating differences in dynamic stability due to amputation from those due to changes in walking speed is therefore critical.

The most direct approach to assess dynamic stability is to impose perturbations that explicitly challenge subjects' ability to walk and then quantify their responses. On irregular surfaces, patients with amputation walk with wider steps, but have exhibited inconsistent results for various measures of "stability" (Lamoth et al., 2010; Curtze et al., 2011; Gates et al., 2012; 2013). Perhaps either the stability measures used in those studies did not adequately capture the effects of the perturbations, or the perturbation paradigms imposed were not sufficiently challenging to elicit appropriate responses. We previously quantified how healthy subjects responded to substantially destabilizing mediolateral perturbations. Subjects exhibited highly significant increases in both kinematic variability (McAndrew et al., 2010) and dynamic instability (McAndrew et al., 2011), validating the idea that local divergence exponents and maximum Floquet multipliers captured these subjects' increased sensitivity to the perturbations imposed. Hak et al. (2013) applied the same perturbation paradigm to patients with transtibial amputation. Although statistically significant, their data show only very small differences between subject groups and perturbation conditions, likely due to methodological differences in how they calculated their local stability exponents.

This study therefore determined how lower limb amputation affects the ability to respond to well-controlled continuous visual or walking surface perturbations. We predicted that persons with transtibial amputations would be (a) more unstable than able-bodied controls during both unperturbed and perturbed walking, (b) would exhibit larger increases in instability than able-bodied controls when subjected to perturbations, and (c) would rely more heavily on visual feedback, resulting in greater increases in dynamic walking instability during visual compared to mechanical perturbations.

2. Methods

Nine persons with unilateral transtibial amputations (TTA) and 13 able-bodied controls (AB) volunteered (Table 1). All TTA subjects walked without assistance using their own prostheses. All TTA patients were active duty military personnel undergoing comprehensive amputee care (Granville and Menetrez, 2010). Participants were screened by a physical therapist to ensure they had no orthopedic or neurological impairments (other than amputation) that might affect their walking. All subjects signed informed consent statements approved by both Brooke Army Medical Center and The University of Texas.

Participants walked in a Computer Assisted Rehabilitation Environment (CAREN) (Motek, Amsterdam, Netherlands). Subjects walked on a 2 m × 3 m instrumented treadmill embedded in a 4 m-diameter six degree-of-freedom motion platform inside a 7 m-diameter dome (Fig. 1A) that created an immersive virtual environment (Sanchez-Vives and Slater, 2005).

The protocol was similar to previous studies (McAndrew et al., 2010, 2011). Each participant walked at a fixed speed scaled to their leg length (l): $v=\sqrt{0.16gl}$ (Vaughan and O'Malley, 2005), where l was defined as distance (m) from the greater

Table 1
Subject characteristics.

	AB ($n=13$)	TTA ($n=9$)
Sex (M/F)	10/3	9/0
Age (years)*	24.8 ± 6.92	30.7 ± 6.75
Height (cm)*	174.8 ± 0.08	176.1 ± 0.11
Body mass (kg)*	79.3 ± 11.56	90.2 ± 16.06
BMI (kg/m^2)*	26.0 ± 3.96	28.86 ± 2.26
Leg length (m)*	0.95 ± 0.05	0.94 ± 0.07
Cause of amputation	---	8 Traumatic 1 Osteosarcoma
Time since amputation (mo)	---	19.8 ± 15.8
Residual limb length (%)[a]	---	55.1 ± 7.47
Avg. prosthetic use (h/day)	---	13.94 ± 2.46
Pain level (#)[b]	---	1.75 ± 1.28

* *Note*: t-tests for group differences: all $p>0.05$.
[a] Residual Limb Length is defined as a percent of length of the lower leg (from the knee to the ankle).
[b] Pain ratings were taken on 10-pt visual analog scale (VAS). All subjects reported pain ≤ 3 at the time of testing.

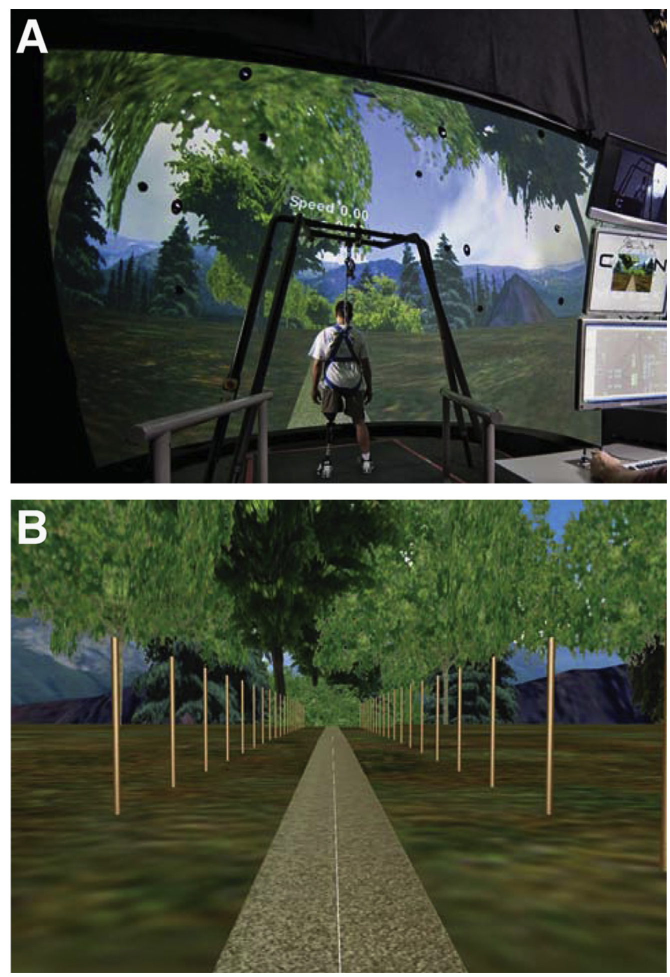

Fig. 1. Experimental setup. A: example photo of a typical person with amputation standing inside the CAREN virtual reality system (Motek, Amsterdam, Netherlands). B: the visual scene used during CAREN trials, depicting a path through a forest with mountains in the background. Both sides of the path were lined with 2.4 m tall white posts spaced every 3 m to increase motion parallax (Bardy et al., 1996; McAndrew et al., 2011).