Techniques to Assess Balance and Mobility in Lower-Limb Prosthesis Users

by

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ABSTRACT

Lower-limb prosthesis users have commonly-recognized deficits in gait and posture control. However, existing methods in balance and mobility analysis fail to provide sufficient sensitivity to detect changes in prosthesis users' postural control and mobility in response to clinical intervention or experimental manipulations and often fail to detect differences between prosthesis users and non-amputee control subjects. This lack of sensitivity limits the ability of clinicians to make informed clinical decisions and presents challenges with insurance reimbursement for comprehensive clinical care and advanced prosthetic devices. These issues have directly impacted clinical care by restricting device options, increasing financial burden on clinics, and limiting support for research and development. This work aims to establish experimental methods and outcome measures that are more sensitive than traditional methods to balance and mobility changes in prosthesis users. Methods and analysis techniques were developed to probe aspects of balance and mobility control that may be specifically impacted by use of a prosthesis and present challenges similar to those experienced in daily life that could improve the detection of balance and mobility changes. Using the framework of cognitive resource allocation and dual-tasking, this work identified unique characteristics of prosthesis users' postural control and developed sensitive measures of gait variability. The results also provide broader insight into dual-task analysis and the motor-cognitive response to demanding conditions. Specifically, this work identified altered motor behavior in prosthesis users and high cognitive demand of using a prosthesis. The residual standard deviation method was developed and demonstrated to be more effective than traditional gait variability measures at detecting the impact of dual-tasking.

Additionally, spectral analysis of the center of pressure while standing identified altered somatosensory control in prosthesis users. These findings provide a new understanding of prosthetic use and new, highly sensitive techniques to assess balance and mobility in prosthesis users.

DEDICATION

This work is dedicated to my parents, who have always encouraged me to achieve my dreams and never let me consider giving anything but my best.

I also dedicate this work to my husband, who despite moving me all over the country in the process, has always given me his unwavering support and never fails to show how proud he is of me becoming a real doctor.

Last, I dedicate this to all my math and science teachers who encouraged my love of learning and put up with the talkative girl who asked all the questions.

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CHAPTER 1

INTRODUCTION

It was estimated that in 2005 over 1 million people in the US were living with lower limb loss, with a major amputation accounting for over half [1]. By 2050 this number is expected to double, primarily due to higher rates of dysvascular disease [1]. Other causes of amputation include trauma, infection, and treatment for bone and joint cancer, along with limb deficiency due to a congenital defect. These non-vascular complications are the leading causes of amputation among younger persons, including military personnel [1, 2]. While the loss of a limb results in a major limitation of mobility, the use of a prosthetic device can restore much of the lost function of the missing limb. However, there is currently no prosthetic device that restores mobility to what is considered unimpaired function. In addition to the mechanical limitations of a prosthetic foot or knee, lower-limb amputees can also experience skin irritation and breakdowns, joint pain, and an increased risk of falls [2-7]. Thus, many amputees express that they experience a reduced quality of life due to their amputation [3, 4, 8].

Research in the area of lower-limb amputation and prosthetic use often works to identify risk factors for reduced quality of life to and improve understanding of prosthetic devices to drive the development of better devices. However, a review of the current state of the field reveals several shortcomings in utility of research practices being employed to characterize prosthetic use [9-14]. Much of the current biomechanics research focuses on standard kinematic and kinetic parameters of amputee gait and posture [15, 16]. Additionally, the research tends to have a strong focus on the prosthetic device and less emphasis on user capabilities and the manner by which they use the prosthetic device.

While this body of work has provided a strong understanding of the mechanics of prosthetic use, current methods fail to provide the sensitivity and specificity needed to identify differences in prosthetic componentry or increase our understanding of the impact of prosthetic use on control of gait and posture beyond basic mechanics [13, 17-21]. Advances in prosthetic design are limited by the lack of information on how prosthesis users are impacted by the device. Additionally, clinicians and payer sources lack measures that effectively distinguish between prosthetic components and assess the effectiveness for different users [13, 17-21]. These issues need to be addressed to improve guidelines for selecting and approving prosthetic prescription and to identify potentially impactful areas of innovation [2, 9-11, 13].

To address these engineering and clinical challenges we must expand our knowledge on prosthetic performance beyond the current scope of the field and develop new measures to assess prosthetic characteristics that consider the prosthesis user response to the device along with the mechanical features. One unaddressed area of exploration is how prosthetic use alters allocation of cognitive resources for motor control. For example, prosthesis users may allocate substantial cognitive resources to achieving reasonable performance while standing or walking, and use of the resources may have implications for behavior in more challenging situations. The evaluation of motor control strategy in prosthesis users goes beyond evaluation of the mechanical impact of the prosthetic devices and examines how prosthesis users adapt behavior to accommodate the mechanical changes imposed by using a prosthesis. The understanding of motor control strategy in other populations has proved useful for developing research protocols and interpreting findings [22-25].

This dissertation examines differences in cognitive resource allocation and motor control strategies in lower-limb prosthesis users during walking and standing to establish experimental methods and outcome measures that are more sensitive than traditional methods to balance and mobility changes. Increased knowledge of the motor control strategy along with the newly developed assessment protocols may be of benefit to both engineers and clinical practitioners in improving prosthetic designs, making evidence based decisions in clinical practice, and providing justification to payer sources.

Areas of impact

Prosthetic design

With the introduction of multiaxial dynamic-response feet and microprocessor knees in the 1990s, the field of lower-limb prosthetics saw major growth in the offerings for lower-limb amputees. As reported on opedge.com as in the fall of 2017, there are more than 13 major companies offering lower-limb products in the US and more abroad. Despite the large number of companies, there is little diversity between their product offerings. For example, most companies offer multiple options for dynamic-response feet that universally incorporate a carbon fiber spring for energy storage and return. Attempts to improve the design include adding elements to provide shock absorption and rotation. However, most studies find little difference between feet within the dynamic-response category or between dynamic response and more traditional feet [11, 12, 26-31]. Additionally, user preference for a type of foot is often mixed and predictors of preference are varied [28, 29, 32, 33]. The abundance of similar products highlights a plateau in design advancement for non-instrumented componentry. There is a need for greater understanding of how users are impacted by their prosthesis outside of standard gait measures in order to identify what elements of prosthetic design change would offer the greatest benefit [2, 13].

Clinical practice

The abundance of similar componentry, along with limited information on functional differences, leaves clinicians few evidence-based guidelines for selecting the best prescription for each individual patient [2, 9, 12, 34, 35]. Since the initial prosthesis is prescribed before patients are able to ambulate, the current extent of the prescription guidelines rely on weight and projected activity level, with many choices for components within these categories [34]. Thus, clinicians rely on their clinical experience, perception of the patient's health and motivation, past experience with a product, and personal preference when selecting prosthetic componentry [9, 35]. Also, it can be difficult to change prescriptions if the user exceeds the projected activity level without measures that capture the improved function. Clinicians need better information about the functional differences between products, but, more importantly, they need improved understanding of their patient's abilities in order to assess which product would best serve individual needs [12, 35].

Similarly, physical therapists working with prosthesis users need better understanding of the user-prosthesis interaction in order to best design a treatment plan for the individual patient. In designing therapy protocols, there are few evidence-based guidelines specific to lower-limb amputees to guide treatment decisions once the patient has received a prosthesis [2]. Improved understanding of how use of a prosthesis uniquely effects motor control strategy may improve therapists' ability to tailor protocols to lower-limb prosthesis users and provide more justification for providing rehabilitation services [15, 36].

Payer sources

The advances in microprocessor and other aspects of prosthetic componentry may improve the mobility afforded by a prosthesis [13]. Users often express a preference for these more advanced components, however in many cases there is insufficient empirical evidence of improved function [10, 13]. As these advances often come with increased cost, the lack of strong evidence to support their benefit limits the justification for approval of higher cost components [12, 13]. Even microprocessor knees, which have been on the market for years and are considered an industry standard, require substantial justification for prescription, often resulting in audits and delayed reimbursement, and are still inaccessible to are large portion of the lower-limb amputee population [13, 37]. A recent market analysis suggests that issues with payment for advanced componentry is one of the primary factors limiting the growth of the prosthetic field [38]. This places strain on clinicians who are trying to balance the burden of providing the best care with the cost of providing the product, while also providing companies with lower reward for developing higher end products [9]. All of which results in reduced benefit to the patient.

The impact of insufficient research that identifies componentry benefits perceived by users was particularly apparent in the 2015 Durable Medical Equipment (DME) and Medicare Administrative Contractor (MAC) release of a joint proposal for large changes to the "Local Coverage Determination and Policy Article" applicable to Medicare reimbursement for lower extremity prosthetics. Medicare billing and reimbursement protocols are accepted as the standard for private insurance companies. One notable change would be removal of billing options for elevated vacuum suspension systems, specifically citing lack of empirical evidence supporting their use. While studies have failed to substantially illustrate the benefits of these systems, their use has become common throughout the prosthetic community due to patient preference and clinician perceived benefits of the suspension style [20, 21]. This highlights one of many examples where lack of sensitivity in research practices are hindering patient access to prosthetic componentry that is strongly supported by clinical observation. Since standard practice in prosthetic research has failed to empirically identify the benefits expressed by users, it is crucial for new protocols to be established that examine factors beyond standard gait analysis and consider other aspects of gait and posture control that could capture and explain the subjective preferences expressed by users.

Understanding motor control in prosthesis users

Motor control strategies determine how people utilize their sensory system to assess their surroundings in relation to their physical condition and allocate cognitive resources between mobility/stability and performance of other concurrent tasks in order to coordinate motor action, such as walking [22, 39]. Many theories have been put forth to explain how cognitive resources are utilized to perform daily activities, which often involve multi-tasking. Many of these suggest sharing of or competition for available resources along with a conscious or unconscious prioritization of certain tasks [39-42]. These theories state that there is a limited amount of resources available for the performance of different tasks and that when demand for those resources increases not all tasks may receive enough resources for best performance [39-42]. This limited resource pool is referred to as the postural reserve, which represents the amount of interference that can be tolerated without detriment to stability [39]. Studies have found that groups with sensorimotor impairments often do not appear to utilize the same resource allocation as healthy controls [23, 25, 39, 43-45]. As lower-limb prosthesis users have an impairment to the sensorimotor system, understanding how cognitive resource allocation is impacted by prosthetic intervention could provide information needed to better understand the complexities of prosthetic use. Further, cognitive resource allocation has not been evaluated in lower-limb prosthesis users.

Observing motor behavior during performance of a single-task

Alterations in motor control in prosthesis users could be due to increased use of the postural reserve or a change in resource allocation. While known gait and posture disruptions in prosthesis users may be suggestive of motor control changes, whether these changes are in part dictated by a change in resource allocation has yet to be determined. It is well established that non-impaired persons perform goal-oriented tasks in a consistent manner and that their performance strategy is dictated by their limb dominance [46, 47]. As amputation puts a constraint on the preferential limb choice, evaluating goal-oriented task performance in lower-limb prosthesis users could provide an indicator of an altered control. For example, if prosthesis users prioritize performance of the goal-oriented task over maintaining the most stable stance it could reflect the role of motivation in the prioritization of resource allocation [39]. Chapter 3 examined limb preference during standing goal-oriented tasks in prosthesis users and non-amputee control subjects.

Insufficient postural reserve may result in an inability to cope with increasing postural demand, such as a destabilizing surface, or impaired stability under simple gait conditions. Individuals with reduced postural reserve are less capable of navigating competing task demands and may be at greater risk for falls [23, 39, 48]. Increasing the demand of a standing or gait task could tax the postural reserve and enhance the need to reallocate cognitive resources to maintain stability [49-51]. Chapters 5 and 6 evaluated postural stability and resource allocation between sensory systems while standing during usual and challenging conditions.

A coordinated gait pattern, considered to be a marker of higher level control of gait [52-54], could be used as an indicator of the availability of the postural reserve during walking. The stride length-cadence relationship provides an indicator of gait coordination [52-54]. Studies in several populations have found that a disruption to this relationship is a strong marker of impaired gait [53, 55, 56]. However, the strength of the stride length-cadence relationship has not been evaluated in the prosthesis user population. Chapters 4 and 8 evaluated the stride length cadence relationship during normal and challenging walking conditions.

Assessing the impact of a concurrent task

Dual-task analysis is often used to evaluate cognitive resource allocation [39-41]. The dual-task paradigm during a gait or posture task introduces an additional cognitive or motor goal to standing or walking [57]. The additional task is believed to compete for the limited resources of the postural reserve [39-41]. Persons with sensorimotor impairments may require greater use of their cognitive-motor resources for standing or walking. Thus, the additional burden may exceed their available resources. Subjects may use specific strategies of resource allocation to best navigate the competing demands to meet their desired goal. For example, individuals who perform the competing tasks without compromising stability, i.e. allocating more resources to systems that help maintain stability, are considered to follow a posture first strategy [23, 40, 44]. Alternatively, individuals may choose to compromise stability in order to achieve the goals of the additional task and are considered to follow a posture second strategy. In addition to reducing the postural reserve due to the sensory and mechanical limitations of the prosthetic device, use of a prosthesis may also increase the cognitive load of standing and walking further increasing the demand on the postural reserve. This may require prosthesis users to alter their resource allocation while dual-tasking.

Dual-tasking has only received limited attention as a research protocol for studying prosthesis users despite its utility in other populations [58], including older adults [59, 60], Alzheimer's disease [61], Parkinson's disease [25], and multiple sclerosis [48] patients. Existing dual-task studies in prosthesis users have focused on above-knee amputees, reported cognitive performance only [13, 62] or found no cognitive-motor interference during walking [51, 63, 64]. Despite these findings, the authors of these studies still maintained that prosthesis users experienced an increased cognitive burden [51, 63]. The lack of evidence supporting the hypothesis may be due to limitations in the analysis method, suggesting the need for a more sensitive measure to detect the impact of dual-tasking. This ambiguity may be due to the selection of sub-optimal outcomes measures or the failure to account for confounding factors, such as velocity changes. A measure that accounts for velocity changes in dual-task gait studies may enhance the utility of dual-task studies in prosthesis users but may also translate to other populations utilizing the paradigm. In Chapter 7 a novel method of gait variability analysis for dual-task studies was developed and evaluated.

Analyzing the level of performance on both task in the dual-task paradigm may provide an improved assessment of dual-task impact and a better characterization of resource distribution [23, 65]. Even greater conclusions can be drawn on self-selected resource allocation if instructed prioritization of resources is also considered. Using measures of dual-task impact to evaluate prosthesis users during walking and standing could reveal if prosthetic use increases the cognitive burden of maintaining stability and alters the how resources are allocated when navigating complex conditions. Chapters 5 and 8 evaluated dual-task impact on posture and gait during challenging standing and walking conditions while also considering the simultaneous performance on a cognitive task.

Statement of purpose

The basic influence of prosthetic componentry on the biomechanics of posture and gait are well documented [12, 16]. However, this level of knowledge is proving insufficient to meet the needs of the field, thus new protocols to evaluate prosthesis users are needed. The purpose of the research, outlined in the Aims below, is to use the framework of the posture reserve and cognitive resource allocation to develop protocols and outcome measures for the field of lower-limb prosthetics that will offer improved means to assess prosthetic use and provide insight into maintenance of stability while standing and walking.

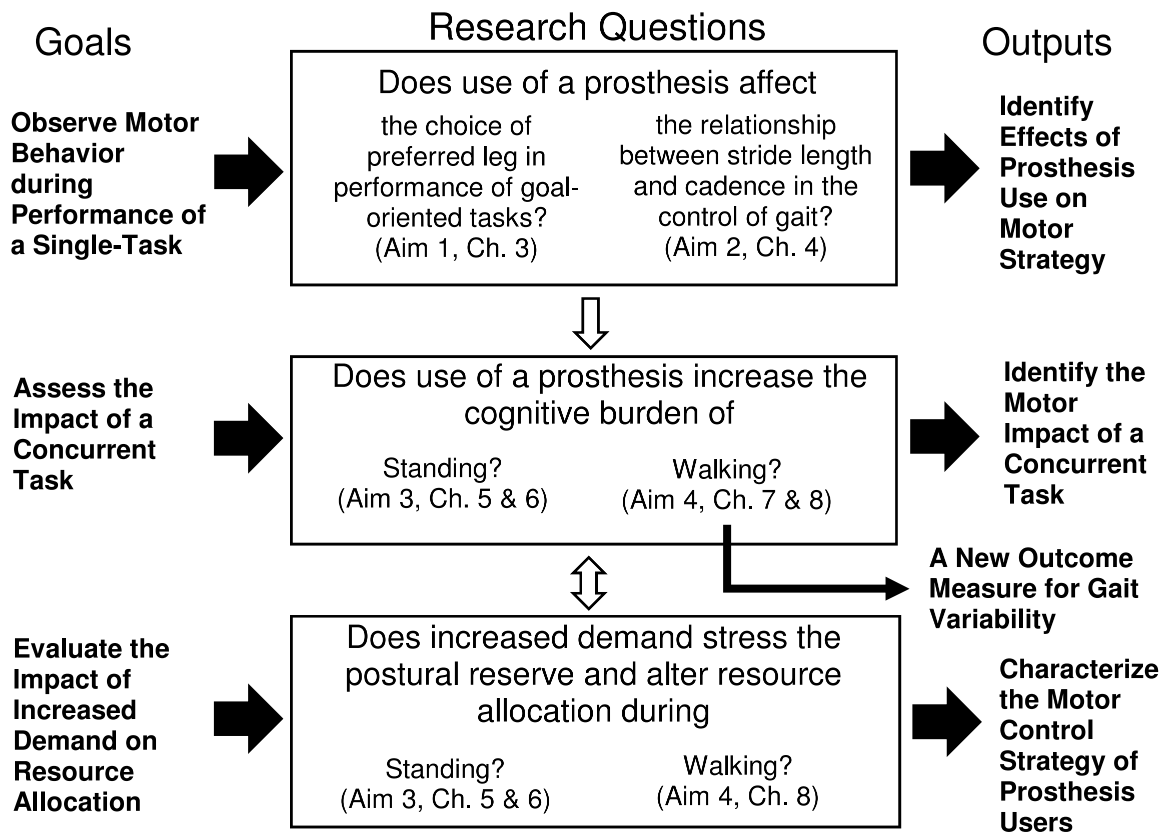

Fig. 1. Framework for addressing the aims of the research question.

Aims

This work aims to document motor behavior during standing and walking during normal and difficult conditions and assess the impact of introducing a concurrent task. Figure 1 provides an overview of the research framework and how each aim will be employed to answer the research questions

Specific Aim 1: Determine if postural goal-oriented motor behavior is altered in prosthesis users (Chapter 3).

Hypothesis 1.1: The prosthetic leg will be the preferred leg in lower-limb prosthesis users as often as the dominant leg in able-body control subjects.

Specific Aim 2: Determine if the neuromotor control mediated relationship between stride length and cadence is disrupted in prosthesis users compared to control subjects (Chapter 4).

Hypothesis 2.1: The linearity of the stride/step length-cadence relationship will be lower in prosthesis users compared with the control subjects.

Hypothesis 2.2: The linearity of the stride length-cadence relationship will discriminate the prosthesis users from control subjects with high sensitivity and specificity.

Specific Aim 3: Evaluate the impact of increased postural challenge and dual-tasking on postural stability during stance in lower-limb prosthesis users (Chapters 5 and 6).

Hypothesis 3.1: Prosthesis users will exhibit higher dual-task impact than control subjects.

Hypothesis 3.2: Greater stability challenge will shift resources toward maintenance of stability and away from performance on the concurrent task.

Hypothesis 3.3: Spectral analysis will identify differences in resource allocation between sensory systems that direct postural control.

Specific Aim 4: Evaluate the impact of increased walking challenge and dual-tasking on gait in lower-limb prosthesis users (Chapters 7 and 8).

Hypothesis 4.1: Prosthesis users will exhibit higher dual-task impact than age- and education-matched control subjects.

Hypothesis 4.2: Greater gait challenge will shift resources toward maintenance of a consistent gait pattern and away from performance of the concurrent task.

Hypothesis 4.3: Accounting for gait velocity in analysis of variability will improve measures of dual-task impact.

These findings will provide new insight into the characteristics of prosthesis users and their use of their postural reserve for resource allocation during standing and walking. Additionally, these studies will introduce new outcome measures aimed at capturing elements of prosthetic use not identified by current practices. These additions to the body of knowledge in the field of lower-limb prosthetics may provide insights that will drive new prosthetic design, improve reimbursement practices, and enable more informed clinical decision making.

References

[1] Ziegler-Graham K, MacKenzie EJ, Ephraim PL, Travison TG, Brookmeyer R. Estimating the prevalence of limb loss in the United States: 2005 to 2050. Arch Phys Med Rehabil. 2008; 89:422-9.

[2] Klute GK, Kantor C, Darrouzet C, Wild H, Wilkinson S, Iveljic S, et al. Lower-limb amputee needs assessments using multistakeholder focus-group approach. J Rehabil Res Dev. 2009; 46:293-304.

[3] Klute GK, Berge JS, Orendurff MS, Williams RM, Czerniecki JM. Prosthetic intervention effects on activity of lower-extremity amputees. Arch Phys Med Rehabil. 2006; 87:717-22.

[4] Sinha R, Van Den Heuvel WJ. A systematic literature review of quality of life in lower limb amputees. Disabil Rehabil. 2011; 33:883-99.

[5] Hunter SW, Batchelor F, Hill KD, Hill A-M, Mackintosh S, Payne M. Risk Factors for Falls in People With a Lower Limb Amputation: A Systematic Review. PMR. 2017; 9:170-80.

[6] Miller WC, Speechley M, Deathe AB. The prevalence and risk factors of falling and fear of falling among lower extremity amputees. Arch Phys Med Rehabil. 2001; 82:1031-7.

[7] Gailey R, Allen K, Castles J, Kucharik J, Roeder M. Review of secondary physical conditions associated with lower-limb amputation and long-term prosthesis use. J Rehabil Res Dev. 2008; 45:15-29.

[8] Legro MW, Reiber G, del Aguila M, Ajax MJ, Boone DA, Larsen JA, et al. Issues of importance reported by persons with lower limb amputations and prostheses. J Rehabil Res Dev. 1999; 36:155-63.

[9] Stark G. Perspectives on how and why feet are presecribed. J Prosthet Orthot. 2005; 17:S18-S22.

[10] Hafner BJ. Clinical prescription and use of prosthetic foot and ankle mechanisms: a review of the literature. J Prosthet Orthot. 2005; 17:S5-S11.

[11] Hafner BJ. Overview of outcome measures for the assessment of prosthetic foot and ankle componets. J Prosthet Orthot. 2006; 18:105.

[12] Klute GK, Kallfelz CF, Czerniecki J. Mechanical properties of prosthetic limbs:Adapting to the patient. J Rehabil Res Dev. 2001; 38:299-307.

[13] Sawers AB, Hafner BJ. Outcomes associated with the use of microprocessor-controlled prosthetic knees among individuals with unilateral transfemoral limb loss: a systematic review. J Rehabil Res Dev. 2013; 50:273-314.

[14] Resnik L, Borgia M. Reliability of outcome measures for people with lower-limb amputations: distinguishing true change from statistical error. Phys Ther. 2011; 91:555-65.

[15] Sawers A, Hahn ME, Kelly VE, Czerniecki J, Kartin D. Beyond componentry: How principles of motor learning can enhance locomotor rehabilitation of individuals with lower limb loss—A review. J Rehabil Res Dev. 2012; 49:1431-42.

[16] Ku PX, Abu Osman NA, Wan Abas WA. Balance control in lower extremity amputees during quiet standing: A systematic review. Gait Posture. 2014; 39:672-82.

[17] Klute GK, Glaister BC, Berge JS. Prosthetic liners for lower limb amputees: a review of the literature. Prosthet Orthot Int. 2010; 34:146-53.

[18] Hofstad C, Linde H, Limbeek J, Postema K. Prescription of prosthetic ankle-foot mechanisms after lower limb amputation. Cochrane Database Syst Rev. 2004; Epub: Feb 20; 10.1002/14651858.CD003978.pub2.

[19] van der Linde H, Hofstad CJ, Geurts AC, Postema K, Geertzen JH, van Limbeek J. A systematic literature review of the effect of different prosthetic components on human functioning with a lower-limb prosthesis. J Rehabil Res Dev. 2004; 41:555-70.

[20] Gholizadeh H, Abu Osman NA, Eshraghi A, Ali S, Razak NA. Transtibial prosthesis suspension systems: systematic review of literature. Clin Biomech. 2014; 29:87-97.

[21] Gholizadeh H, Abu Osman NA, Eshraghi A, Ali S. Transfemoral prosthesis suspension systems: a systematic review of the literature. Am J Phys Med Rehabil. 2014; 93:809-23.

[22] Yogev-Seligmann G, Hausdorff JM, Giladi N. The role of executive function and attention in gait. Mov Disord. 2008; 23:329-42.

[23] Yogev-Seligmann G, Rotem-Galili Y, Dickstein R, Giladi N, Hausdorff JM. Effects of explicit prioritization on dual task walking in patients with Parkinson's disease. Gait Posture. 2012; 35:641-6.

[24] McCulloch KL, Buxton E, Hackney J, Lowers S. Balance, attention, and dual-task performance during walking after brain injury: associations with falls history. J Head Trauma Rehabil. 2010; 25:155-63.

[25] Bloem BR, Grimbergen YA, van Dijk JG, Munneke M. The "posture second" strategy: a review of wrong priorities in Parkinson's disease. J Neurol Sci. 2006; 248:196-204.

[26] Hansen AH. Scientific Methods to determine functional performance of prosthetic ankle-foot systems. J Prosthet Orthot. 2005; 17:S23-S9.

[27] Segal AD, Orendurff MS, Czerniecki JM, Shofer JB, Klute GK. Local dynamic stability of amputees wearing a torsion adapter compared to a rigid adapter during straight-line and turning gait. J Biomech. 2010; 43:2798-803.

[28] Klodd E, Hansen AH, Fatone S, Edwards M. Effects of prosthetic foot forefoot flexibility on oxygen cost and subjective preference rankngs of unilateral transtibial prosthesis users. J Rehabil Res Dev. 2010; 47:543-52.

[29] McMulkin ML, Osebold WR, Mildes RD, Rosenquist RS. Comparison of three pediatric prosthetic feet during functional activities. J Prosthet Orthot. 2004; 16:78-84.

[30] Berge J, Czerniecki J, Klute GK. Efficacy of shock-absorbing versus rigid pylons for impact reduction in transtibial amputees based on laboratory, field, and outcome metrics. J Rehabil Res Dev. 2005; 42:795-808.

[31] Segal AD, Orendurff MS, Czerniecki JM, Shofer JB, Klute GK. Transtibial amputee joint rotation moments during straight-line walking and a common turning task with and without a torsion adapter. J Rehabil Res Dev. 2009; 46:375-83.

[32] Hafner BJ, Sanders JE, Czerniecki J, Fergason J. Energy storage and return prostheses: does patient perception correlate with biomechanical analysis? Clin Biomech. 2002; 17:325-44.

[33] Delussu AS, Brunelli S, Paradisi F, Iosa M, Pellegrini R, Zenardi D, et al. Assessment of the effects of carbon fiber and bionic foot during overground and treadmill walking in transtibial amputees. Gait Posture. 2013; 38:876-82.

[34] Shurr D. Clinical perspectives on the prescription of prosthetic foot-ankle mechanisms. J Prosthet Orthot. 2005; 17:S31-S2.

[35] Borrenpohl D, Kaluf B, Major MJ. Survey of U.S. Practitioners on the Validity of the Medicare Functional Classification Level System and Utility of Clinical Outcome Measures for Aiding K-Level Assignment. Arch Phys Med Rehabil. 2016; 97:1053-63.

[36] Huang HJ, Mercer VS. Dual-task methodology: applications in studies of cognitive and motor performance in adults and children. Pediatr Phys Ther. 2001; 13:133-40.

[37] Hafner BJ, Smith DG. Differences in function and safety between Medicare Functional Classification Level-2 and-3 transfemoral amputees and influence of prosthetic knee joint control. J Rehabil Res Dev. 2009; 46:417-33.

[38] McGimpsey G, Bradford TC. Limb prosthetics services and devices. Worcester, MA: Bioengineering Institute Center for Neuroprosthetics, Worcester Polytechnic Institution; 2010.

[39] Yogev-Seligmann G, Hausdorff JM, Giladi N. Do we always prioritize balance when walking? Towards an integrated model of task prioritization. Mov Disord. 2012; 27:765-70.

[40] Lacour M, Bernard-Demanze L, Dumitrescu M. Posture control, aging, and attention resources: Models and posture-analysis methods. Clin Neurophysiol. 2008; 38:411-21.

[41] Beurskens R, Steinberg F, Antoniewicz F, Wolff W, Granacher U. Neural correlates of dual-task walking: Effects of cognitive versus motor interference in young adults. Neural Plast. 2016; 2016:1-9.

[42] Bonnet CT, Baudry S. Active vision task and postural control in healthy, young adults: Synergy and probably not duality. Gait Posture. 2016; 48:57-63.

[43] Kelly VE, Schrager MA, Price R, Ferrucci L, Shumway-Cook A. Age-associated effects of a concurrent cognitive task on gait speed and stability during narrow-base walking. J Geront A. 2008; 63:1329-34.

[44] Shumway-Cook A, Woollacott M, Kerns KA, Baldwin M. The effects of two types of cognitive tasks on postural stability in older adults with and without a history of falls. J Geront A. 1997; 52:232-40.

[45] Swanenburg J, de Bruin ED, Favero K, Uebelhart D, Mulder T. The reliability of postural balance measures in single and dual tasking in elderly fallers and non-fallers. BMC Musculoskelet Disord. 2008; 9:162.

[46] Schneiders AG, Sullivan J, O'Malley KJ, Clarke SV, Knappstein SA, Taylor LJ. A valid and reliable clinical determination of footedness. PMR. 2010; 2:835-41.

[47] Sadeghi H, Allard P, Prince F, Labelle H. Symmetry and limb dominance in able-bodied gait: a review. Gait Posture. 2000; 12:34-45.

[48] Etemadi Y. Dual task cost of cognition is related to fall risk in patients with multiple sclerosis: a prospective study. Clin Rehabil. 2017; 31:278-84.

[49] Vanicek N, Strike S, McNaughton L, Polman R. Postural responses to dynamic perturbations in amputee fallers versus nonfallers: a comparative study with able-bodied subjects. Arch Phys Med Rehabil. 2009; 90:1018-25.

[50] Mohieldin A, Chidambaram A, Sabapathivinayagam R, Al Busairi W. Quantitative assessment of postural stability and balance between persons with lower limb amputation and normal subjects by using dynamic posturography. Maced J Med Sci. 2010; 3:138-43.

[51] Morgan SJ, Hafner BJ, Kelly VE. Dual-task walking over a compliant foam surface: A comparison of people with transfemoral amputation and controls. Gait Posture. 2017; 58:41-5.

[52] Egerton T, Danoudis M, Huxham F, Iansek R. Central gait control mechanisms and the stride length - cadence relationship. Gait Posture. 2011; 34:178-82.

[53] Morris M, Iansek R, Matyas T, Summers J. Abnormalities in stride length-cadence relation in parkinsonian gait. Mov Disord. 1998; 13:61-9.

[54] Zijlstra W, Rutgers AWF, Hof AL, Van Weerden TW. Voluntary and involuntary adaptation of walking to temporal and spatial constraints. Gait Posture. 1995; 3:13-8.

[55] Barak Y, Wagenaar RC, Holt KG. Gait characteristics of elderly people with a history of falls: a dynamic approach. Physical Therapy. 2006; 86:1501-10.

[56] Callisaya ML, Blizzard L, McGinley JL, Srikanth VK. Risk of falls in older people during fast-walking - The TASCOG study. Gait Posture. 2012; 36:510-5.

[57] McIsaac TL, Lamberg EM, Muratori LM. Building a framework for a dual task taxonomy. Biomed Res Int. 2015; 2015:1-10.

[58] Fritz NE, Cheek FM, Nichols-Larsen DS. Motor-Cognitive Dual-Task Training in Persons With Neurologic Disorders: A Systematic Review. J Neurol Phys Ther. 2015; 39:142-53.

[59] Amboni M, Barone P, Hausdorff JM. Cognitive contributions to gait and falls: evidence and implications. Mov Disord. 2013; 28:1520-33.

[60] Beauchet O, Annweiler C, Allali G, Berrut G, Dubost V. Dual task-related changes in gait performance in older adults: a new way of predicting recurrent falls? J Am Geriatr Soc. 2008; 56:181-2.

[61] Muir SW, Speechley M, Wells J, Borrie M, Gopaul K, Montero-Odasso M. Gait assessment in mild cognitive impairment and Alzheimer's disease: The effect of dual-task challenges across the cognitive spectrum. Gait Posture. 2012; 35:96-100.

[62] Williams RM, Turner AP, Orendurff M, Segal AD, Klute GK, Pecoraro J, et al. Does having a computerized prosthetic knee influence cognitive performance during amputee walking? Arch Phys Med Rehabil. 2006; 87:989-94.

[63] Morgan SJ, Hafner BJ, Kelly VE. The effects of a concurrent task on walking in persons with transfemoral amputation compared to persons without limb loss. Prosthet Orthot Int. 2016; 40:490-6.

[64] Lamoth CJ, Ainsworth E, Polomski W, Houdijk H. Variability and stability analysis of walking of transfemoral amputees. Med Eng Phys. 2010; 32:1009-14.

[65] Plummer P, Eskes G. Measuring treatment effects on dual-task performance: a framework for research and clinical practice. Front Hum Neurosci. 2015; 9:225.

CHAPTER 2

BACKGROUND LITERATURE

Amputee gait and posture

It is well established that both above- and below-knee prosthesis users have altered gait and posture mechanics [1-5]. The loss of active control of the missing joints, reduced propulsion and braking from musculature, and loss of sensory feedback place limitations on prosthesis users' mobility. Advances in prosthetic componentry attempt to restore some of these lost functions, however, characteristics of the amputee, such as strength, limb health, and comorbidities, play a major role in successful prosthetic use and restoration of mobility [5, 6].

The majority of lower-limb prosthetic components are passive devices that provide a limited range of motion to replace the lost joint movement. Some products on the market offer various levels of controlled resistance through hydraulic and/or microprocessor features. Additionally, some microprocessor componentry allows for active motion to better accommodate different surfaces and conditions. However, very few of these products offer powered prolusion and those that do see limited public use [7, 8]. These products restore some level of mobility to prosthesis users, yet no current products provide motion, braking, or propulsion that is controlled by the user and none of them restore sensory feedback. Thus, there are still many gait and posture deviations common among lower-limb prosthesis users even when using advanced componentry [9-16].

Prosthesis users tend to show signs of greater instability while standing, particularly under more challenging conditions, compared to non-amputees [17, 18].

Studies find that the indicators of instability, such as increased COP movement, are related to higher level of amputation, shorter residual limb length, amputation due to vascular complications, older age, and the presence of comorbidities [2, 4, 19, 20]. As standing balance is greatly influenced by sensory feedback and active control at the ankle [21], it is not surprising that prosthesis users have greater postural instability than non-amputees [19, 22-24]. Many users compensate for the limited utility of the prosthetic side by placing greater weight and reliance on the sound side [20]. This overuse is considered a main cause of the frequent occurrence of musculoskeletal issues on the sound side [25].

Many features of gait are also regularly reported to be altered in lower-limb prosthesis users. Commonly reported temporal-spatial features include slower self-selected walking speed and asymmetry in many parameters resulting in more time spent on the sound side [5, 26-28]. These characteristics have been reported for all amputation levels, but are often more pronounced in above-knee prosthesis users [5, 28]. The use of microprocessor componentry has potential to reduce many of these abnormalities [29], however findings are often inconsistent [30].

Joint kinematics and muscle activation are also altered during prosthetic gait [31]. The passive action of the prosthetic foot results in less motion at the prosthetic ankle [5]. Even in below-knee prosthesis users with an intact knee, the first knee flexion peak after initial contact is delayed and is much smaller than in non-amputees. The initial knee flexion peak is almost nonexistent in above-knee amputees. The sound side also exhibits joint motion differences. To increase stability during gait, amputees produce stronger and more sustained muscle contractions. All of these features are more exaggerated under complex gait conditions such as ramp or stair ambulation [5, 32].

Despite the substantial knowledge of gait and posture characteristics common to prosthesis users that has been described in the literature, this information has proved insufficient in providing protocols sensitive enough to detect the impact of altering prosthetic prescription [9, 10, 33-36]. Studies examining different styles of prosthetic components often fail to report substantial differences between designs despite subject preference [13, 30, 35, 37-42]. The lack of sensitivity of current protocols has resulted in limitations to device design, evidence-based clinical practice, and insurance approval [42, 43]. These shortcomings suggest that further research is needed that examines prosthesis users beyond standard gait analysis [42, 44-46].

While the gait and standing characteristics of prosthesis users are well documented and many features are explained by limitations of prosthetic devices, the impact of prosthetic use on the neurocontrol of mobility and stability has yet to be thoroughly examined. Understanding how use of the prosthetic devices alters control of gait and posture could provide insight into the needs of prosthesis users, provide clinicians with a better understanding of patients, and help develop more sensitive measures for testing prosthetic componentry [46].

Models of neurocontrol

Postural control, utilized for standing and walking, is actively and passively maintained through coordinated responses to visual, vestibular, and somatosensory sensory feedback loops along with mechanical support provided by the musculoskeletal system and cognitive engagement [47, 48]. Many theories suggest that these mechanisms work together through a shared use of a limited supply of cognitive resources [47-51]. Environmental conditions or physical limitations, such as low lighting or vestibular dysfunction, or multi-tasking may increase the demand for the available resources [47-49]. If demand for resources exceeds availability, a decline in performance may occur [47]. Many studies illustrate detriments in balance control when feedback loops are altered or inhibited and show that persons with no other limitations are better able to maintain their normal posture and gait characteristics when resource demand increases [47, 52-55]. Those with no impairments are presumed to have more of these resources available to distribute amongst the demands of the different tasks. The summation of these resources and the ability to utilize them as need arises is often referred to as the postural reserve [49].

Postural reserve is defined by Yogev et al. as "the individual's capability to respond most effectively to a postural threat [49]." Stated differently, it reflects the amount of resources available that can be utilized through the activation of postural control mechanisms to respond to the postural demand. Thus, persons with a larger postural reserve can allot more resources or greater attention to other tasks while still providing enough resources to maintain stability. Postural reserve is affected by several factors including strength, sensory feedback capacity, motor control capabilities, and cognitive processing ability [49]. Amputation disrupts many of these factors, which presumably reduces available resources and requires greater activation of other control systems in the postural reserve [23]. A reduced postural reserve may require prosthesis users to place less priority on non-stability related tasks than non-impaired subjects [48, 49].

The cross-domain competition model [48] and the similar central capacity sharing model [50, 56] propose that the competing activities must share resources and under conditions where demand exceeds available resources, performance inevitably declines [47]. Support for these models comes from studies demonstrating a decline in performance while carrying out concurrent tasks while standing or walking, referred to as dual-tasking. Further, researchers have found changes neural activation corresponding decrements in gait and posture performance while dual-tasking, supportive of models suggesting a shortage of resources [50, 57]. While these models provide an explanation for many findings, some argue they fail to explain dual-task findings that report no change in dual-task performance or improvement in performance [48, 58]. However, Tombu and Jolicoeur argue that the central capacity sharing model can still account for these findings by suggesting that not all systems share the same pool of resources or that persons may not always utilize all available resources [56].

Within the competition or sharing models there is some suggestion of the ability to direct resource allocation to the task considered most important [56]. The task prioritization model expands on theories of resource competition and offers a more comprehensive means of capturing and explaining the many possible outcomes in multi-tasking behavior [48, 49]. It suggests that individuals can coordinate resource allocation between multiple tasks based on personal priority of various factors, such as necessity of the non-stability task and physical safety [49].

The posture first strategy has been suggested as one common behavior supportive of a prioritization model. It suggests that people maintain stability by prioritizing resource allocation towards the gait or posture tasks when conflicting tasks require the use of overlapping systems. The posture first strategy was first proposed by Shumway-Cook in 1997 [54]. Many studies have found that while unimpaired adults tend to follow this strategy, persons with various impairments do not, which may increase instability and fall risk. Studies have found that groups with cognitive-motor impairments, such as Parkinson's disease, allow performing a concurrent task while standing or walking to disrupt stability more than unimpaired subjects [47-49, 59]. This is suggested to be counter to the posture first strategy, as following the posture first strategy would distribute resources in such a way that stability would not be compromised. While the posture first strategy focuses on the prioritization of stability, other factors such as mood, personality, risk acceptance/aversion, and nature of the additional tasks could also be factors in the prioritization of resource allocation [49]. Within the competition models the prioritization model can provide some explanation for dual-task behavior that does not follow the expected pattern of performance decline [54, 60].

Another explanation of dual-task performance that results in no change or improvement in both tasks comes from the level of alertness hypothesis [51, 52]. When a person has all resources available, low-demand competing tasks do not exceed the available reserve and do not pose such a threat as to warrant allocating additional resources to performance. However, when demand increases, posing a greater threat, additional resources can be engaged [56]. Wrightson et al. provides neurological evidence that individuals may allow some decrements to performance while dual-tasking despite further resources being available; and when those resources are activated, gait performance can return to near normal characteristics [61].

In clinical research studies, neurocontrol models have been most widely investigated in populations with a neurological impairment [49, 53, 54, 59, 62, 63]. In these populations, such as stroke or Parkinson's disease, the impairment is typically to central nervous system dysfunction, which may impact cognitive or neuromotor function. Examinations of resource allocation in these populations are used to assess the impact that the central neurological impairment has on the ability to dual-task while safely maintaining control of gait or posture. However the study of resource allocation in prosthesis users would differ from the majority of previous research in the field as the primary impairment is in the periphery. Thus, the rationale for examining resource allocation in prosthesis users is not to assess the impact of a central nervous system impairment on lower functions but to evaluate how a peripheral impairment, i.e. amputation and use of a prosthetic device, increases the burden on the higher neurological systems by altering the utilization of resources in the postural reserve. Investigation of the postural reserve and how prosthesis users prioritize competing tasks could provide greater understanding of stability control in prosthesis users.

Dual-task research background

Concurrent tasks

A dual-task paradigm is used to test theories of neurocontrol during multi-tasking. Dual-tasking involves the performance of two (or more) concurrent tasks. In studies investigating the impact of multi-tasking on stability control, one of the concurrent tasks is usually standing or walking. The additional concurrent task can vary widely, however, they typically fall into the category of a cognitive or an additional motor task [64]. These additional tasks increase the complexity of performance by inducing a separate measurable goal to the standing or walking task [64]. Cognitive tasks have been used in many studies reported in the dual-task literature. Common cognitive tasks include, serial subtraction [65-67], backwards spelling [68], verbal fluency (listing words in a specific category) [69, 70], and different variations on the Stroop test [55, 71, 72]. Cognitive task performance often involves the subject providing verbal responses. However, some studies may use responses that require an action such as pressing a button [57], common with an auditory Stroop test, or silent mental performance [73]. Some studies also utilize tasks that requires subjects to respond to a visual cue [74].

Motor tasks used in dual-task studies are more diverse [75], however a key component of the tasks are the motor action should be independent of the primary motor task (i.e. standing or walking) [64]. For example, buttoning a shirt while standing would be considered a dual-task while transporting an object would not. However, this definition is not universally used [75, 76]. Often motor tasks are chosen to represent real life multi-tasking activities [75, 77]. While a motor task typically does not have a verbal response, it can require visual attention or responding to a visual or auditory cue [78, 79]. Thus, within both task options the concurrent task can involve a combination of demands from visual, auditory, verbal, cognitive, and motor resources.

Dual-task difficulty

In addition to utilization of a concurrent cognitive or motor task, researchers also attempt to probe resource allocation and dual-task behavior by increasing the difficulty of the single-task standing or walking [64, 80]. The increased difficulty aims to further increase the burden of standing or walking, making subjects more susceptible to the dual-task interference. Means of increased difficulty include eyes closed conditions [81], destabilizing surfaces [80-82], narrow or complex walkways [73, 83, 84], and increasing task novelty or complexity [64]. There are mixed reports of these methods inducing a change in the dual-task response [80, 82].

Dual-task outcomes

Measures of dual-task impact are diverse but often focus on outcomes associated with predicting instability. The most commonly reported outcomes in gait dual-task studies are changes in gait speed and variability of temporal-spatial gait parameters [85, 86]. Changes in gait characteristics while dual-tasking have been associated with fall risk and instability [87-89]. In postural control studies, outcomes typically focus on measures of increased center of pressure (CoP) movement. For example, Sample et al. found that an increase in sway area and medial-lateral amplitude during a motor dual-task differentiated between older adult fallers and non-fallers [90]. Other studies have identified dual-task impact on sway velocity and path length [80, 91]. While traditional measures of CoP movement are common in studies of dual-task postural control, Lacour et al. argues against drawing strong conclusions of postural stability from these measures, since a decrease in postural sway can reflect a stiffening strategy, often associated with a fear of falling, rather than improved control [48]. Alternatively, several studies have highlighted the utility of non-linear or spectral analysis of the CoP signal in dual-task studies. For example Collins et al. [92] and Ghulyan et al. [93] both identified that spectral analysis of the CoP during single-task standing better differentiated between younger and older subjects than traditional measures. Bernard-Demanze et al. also reported better detection of postural changes due to dual-tasking using spectral analysis [80]. Sample reported similar utility for evaluation of the impact of cognitive but not motor dual-task [90].

In addition to evaluation of dual-task performance on gait and posture characteristics, Plummer and Eskes highlight the importance of evaluating changes in both tasks to fully appreciate dual-task behavior [60]. A full picture of resource allocation requires evaluation of the interplay between each task. When both tasks are considered, there are 9 potential outcomes that can be visualized as a 3x3 matrix of facilitation, no interference, or interference for each task which each region offering a potential resource allocation interpretation. For example, improvement or no change in gait or posture performance is often interpreted as prioritization of stability/mobility, however simultaneous improvement on the concurrent task shows mutual facilitation [60], more in line with increased resource activation or the level of alertness hypothesis [51, 52].

Dual-task analysis in prosthesis users

Geurts et al. evaluated the impact of a cognitive task on postural control in lower-limb prosthesis users before and after rehabilitation [72, 94]. These studies found greater dual-task interference in prosthesis users than non-amputee control subjects at both time points, however the effect was reduced after rehabilitation training [72, 94]. In contrast, several more recent studies on posture [20] and gait [3, 82, 95] dual-task analysis has not identified greater disruptions to stability or mobility while performing a concurrent task. Other studies evaluating prosthesis users have only evaluated the cognitive performance of dual-tasking, but also reported no impact [30, 70]. Overall, dual-tasking has seen little use in the prosthesis user population and the majority of studies have focused on evaluating above-knee prosthesis users. These findings suggest a need for more sensitive evaluations of dual-task performance in prosthesis users to increase understanding of cognitive resource allocation in response to prosthetic use.

Spectral analysis of the center of pressure signal

Frequency components of postural control

It is long established that 1 Hz marks a boundary in the power spectrum of human postural control [96, 97]; this has been repeatedly supported [98-103] as research in the field has progressed. Many of these later studies worked to understand the physiological role or mechanisms that give rise to this separation. For example, Diener et al. has suggested that spinal reflex control, such as the Golgi tendon organs and spindle afferents, can sufficiently respond to high frequency (>1Hz) perturbations and that postural control requiring higher cortical processing respond to lower frequency (<1 Hz) changes [98, 99].

However, it has been argued that higher cortical processing, i.e. active control, plays little role in postural control during quiet, unperturbed stance and the primary mechanism is joint stiffening in response to sensory reflexes [104, 105]. Winter et al. suggests that a near 0^{th} order system is the primary driver of postural control because of nearly in phase movement between center of mass and CoP that would have a longer response time due to afferent and efferent conduction delays and/or higher order system dynamics if movements were directed by a higher order system [105]. The authors further suggest that the ability to respond to perturbations below sensory thresholds also support a passive or feedforward control system [105]. Morasso et al. argues against the simplicity of the control system model proposed by Winter et al [106, 107]; pointing out that a postural control model that simply relies on joint stiffening ignores a multitude of evidence pointing to the importance of sensory systems in postural control. Instead, Morasso and Sanguineti suggest a model where approximately 60% of postural control is due to muscle/joint stiffness (feedforward or open-loop) and 40% to active control mediated by sensory systems (feedback or closed-loop) [107]. Further, even if the stiffening model proposed by Winter et al. is true for only quiet standing, models that incorporate feedforward and feedback control may be more appropriate for evaluating postural control during functional standing.

Collins and De Luca and later Singh et al. used different methods of critical point detection in their assessment of postural control and further confirmed 1 Hz as the general transition point between open and closed loop control systems [100, 102]. Critical point detection identifies a distinct time point in the COP signal where the characteristics of the signal changes, establishing a point where the closed loop system takes control of posture mechanics to direct the body back to equilibrium. While 1 Hz has become a general standard for transition between the two control systems, there are across-subject differences in the specific transition point with one study identifying values that ranged from 0.33 to 1.67 s (mean 1 s) in the AP direction and 0.81 to 1.30 s (mean 1 s) in the ML direction [100] and the critical point occurring around 0.62s in another [102]. Open-loop control is suggested to allow sway to 'drift' until closed loop control takes over. Thus, the use of an open-loop control strategy simplifies the amount of high level processing for postural control [100], fills in the gaps in feedback loop delays [101], and reduces energy expenditure in maintaining upright stance [102]. Singh et al. found that under more challenging standing conditions closed-loop control utilization is increased in order to establish more active control of posture and reduce the risk to stability [102]. However, it should be noted that in contrast to the models proposed by both Winter and Morasso, others believe that open-loop (or feedforward) control does not play a strong role in postural control and that closed-loop feedback is sufficient for postural control [108].

Within the low frequency range associated with closed-loop control, many studies have identified distinct peak frequencies or frequency bands [97, 103, 109]. With the knowledge that the visual, vestibular, and somatosensory systems are the primary contributors to postural control and they require higher level cortical processing that create longer feedback loops, it has been proposed that these frequency bands reflect each system's contribution to postural control [98, 99, 103]. Many different approaches have been used to define these frequency bands and assign them to a specific sensory system. Different methods include studying populations with impaired or absent sensory systems [110-112], limiting the contribution of sensory systems through experimental manipulation [48, 80, 113-116], or applying postural perturbations at specific frequencies [98, 99]. For example, Diener et al. found that subjects had a delayed vestibular response (postural corrections to prevent falling while eyes were closed) to perturbations delivered below 0.3 Hz, leading the authors to suggest that the vestibular system does not contribute to postural control at frequencies below that frequency [99]. In studying patients with and without peripheral neuropathy, Oppenheim et al. reported that only changes in power for the frequency band between 0.5 and 1 Hz distinguished between patient groups, assigning this band to somatosensory control [110]. Collectively, these studies have produced a general consensus on the frequency bands and the associated sensory systems: vision is associated with very low frequencies (< 0.1 Hz), vestibular low frequencies (~ 0.1-0.5 Hz), somatosensory middle frequencies (~ 0.5-1 Hz), and feedforward or open-loop is associated with high frequencies (> 1 Hz). While these bands are widely reported, differences between studies and reported within subject variability suggest these are general guides without firm boundaries. However, analysis of the sensory systems associated with specific frequency bands through spectral analysis could provide increased understanding of resource allocation between systems.

Spectral analysis

Several researches have provided evidence that spectral analysis of the CoP signal is more sensitive than traditional measures in detecting changes in postural control as the result of dual-tasking or different standing conditions or differentiating between groups [80, 90, 92, 93, 114, 117]. Within spectral analysis methods, wavelet analysis has been suggested to be particularly well suited for the analysis of standing CoP signals [48, 114, 115, 118, 119]. Wavelet analysis uses variable-sized, time-scale specific windows to deconstruct the signal into time-scale bands; the time-scales can then be transformed to frequencies. Specifically, a mother wavelet, a time and frequency localized function with a mean of zero, is compared to a section of the signal being analyzed and the correlation between the two signals is calculated. The process is repeated as the wavelet is shifted along the signal. The wavelet is then scaled, stretched or compressed, and compared to the length of the signal again. This process is repeated for each scale. The scale represents for coarseness of the comparison to the signal and is inversely related to frequency. A high scale results in a more stretched wavelet and captures low frequency elements of the signal. The summation of the correlations for each scale represent the energy content of the signal in a specific frequency band [120-122].

Wavelet analysis is favored for the evaluation of time-varying, non-stationary signals and is superior to Fourier analysis at characterizing the spectral power in non-dominant frequencies [114, 115, 120-122]. These features are useful for analysis of the CoP movement as the signal has been reported as being non-stationary and is typically dominated by low frequency energy [114, 115]. While popularity of the analysis method has grown in recent years, wavelet analysis suffers from a lack of guidelines directing its use for CoP evaluation, primarily in the selection of the mother wavelet for signal decomposition [120].

Analysis of resource allocation in non-dual-task paradigm

Altered control mechanisms and prioritization during standing tasks: Amputation vs. natural laterality

While dual-tasking is the most common means of assessing resource allocation, the implications of allocation apply to aspects of stability and mobility outside the factors captured by dual-task analysis. Prioritization between maintaining stability and successful completion of the competing task plays a major role in understanding resource allocation. While the limits of the person's postural reserve and assessment of their own abilities are consistent factors, other elements such as desire to complete the task play a role in prioritization [49]. The element of choice is particularly important when executing goal-oriented tasks. Goal-oriented tasks involve performing an action with a defined purpose and clear indicators of success or failure [64], such as stomping on a bug. The desire to complete the goal may determine the risk to stability that the person may deem acceptable. For example, a person particularly bothered by the presence of an insect may allow a risk to stability beyond their usual level.

In standing, goal-oriented tasks which rely on stabilizing with one foot and action with the other, the means of execution are typically mediated by natural lateralization.

Lateralization is the preferential use of one side of the body over the other for performance of various tasks, and is also referred to as limb dominance [123]. The unilateral preference for one side of the body is associated with dominance of cranial hemispheres. In fact, while cerebral dominance is often associated with hand preference, a pair of studies by Elias et al. found that foot preference is a better predictor of cerebral lateralization [124, 125]. In postural tasks, lateralization determines which leg is used for stabilization and which is used to perform the task. The dominant leg is typically used to perform the action of the task, such as kicking a ball, while the non-dominant leg is used for stabilization. The leg used to perform the action is called the preferred leg [123].

Amputation places constraints on limb selection potentially altering the natural control mechanisms used to direct task performance [126]. With the natural lateralization disrupted, the choice of action or stabilizing leg is now a more conscious choice. This choice is most likely mediated many factors including the risk to stability and personal motivation to complete the task [49]. These factors are weighted between the concurrent objectives of maximizing reward (successful task completion) and minimizing risk (maintain balance). For some tasks, choosing to balance on the prosthetic side may allow the controlled articulation of the intact side to better manipulate the task, despite a potential risk to stability. Amputation causes even simple tasks to require prioritization of performance vs. stability from prosthesis users while non-amputees simply rely on their natural dominance. Knowledge of how prosthesis users choose to navigate the competing demands of various postural tasks can be used to begin to assess how motivation to perform a goal may direct resource allocation.

The role of the dominant limb also has implications for gait control such as normal walking [127, 128], gait initiation [129], turning [126, 130], and stair ambulation [131]. The studies supporting a relation between laterality and gait mechanics report the presence of a supporting limb and propulsive (action) limb that are specific to a side of the body or the reported dominant limb [128]. Thus, increased understanding of the impact of amputation on limb preference during standing tasks could lead to improved understanding of gait control.

Indicator of decreased postural reserve: Stride-length cadence relationship

In order to understand resource allocation in lower-limb prosthesis users, the impact of prosthetic use on the postural reserve should be characterized. Signs of altered control of gait mechanics linked to instability could serve as indicators of decrement in the postural reserve. While the observed gait changes due to prosthetic use are well documented, there have been few studies on the underlying neuromotor control aspects of these changes. One measure that attempts to quantify and understand the neuromotor control mechanisms that direct gait is the stride length-cadence linear relationship [132, 133]. Studies have shown that healthy adults modulate stride length and cadence in such a way that these parameters tightly co-vary over a range of normal walking speeds [132, 134-137]. This relationship can be reported using a measure of linearity, such as the goodness of linear fit of the regression line (R^2) [132, 136]. This coupled relationship, with similar slopes across the population, creates a predictable universal pattern for how speed is modulated in healthy adults. It also provides a single measure of gait control that incorporates a wide range of walking speeds and combines two commonly reported gait parameters. Therefore, it provides a more simplistic measure of coordinated gait that does not rely on interpretation of multiple variables.

While the existence of the stride length-cadence relationship has long been documented [134, 135, 137, 138], only more recent studies have examined the neuromotor control implications of the relationship [132, 136]. These studies have reported that when either stride length or cadence is restricted, such as asking someone to walk in rhythm to a specific beat, the variables begin to decouple. These studies suggest that the linear stride length-cadence relationship represents an automatic neuromotor control mechanism and, when faced with a restriction, more conscious and less efficient control mechanisms are activated. Thus, it is suggested that deterioration of this relationship may represent lower gait quality [132, 136].

By quantifying a high-level control mechanism representative of stable gait, encompassing two commonly reported gait parameters, and examining multiple speeds all in one measure, the stride length-cadence relationship serves as a good model for predicting the state of a person's postural reserve. Thus, a decrease in the linearity of the relationship between stride length and cadence in amputees compared to non-amputees could signify a more conscious control of gait resulting in a decrease in the resources available for more complex gait tasks. Additionally, if a weaker relationship also signifies activation of more conscious gait control mechanisms, it would suggest that amputees have additional cognitive burden placed on them even during the simplest walking conditions.

There are important mathematical considerations when calculating goodness of linear fit and making comparisons between groups or studies. These considerations involve the number of data points and the range the data (i.e. speeds) being analyzed, as both can have an impact on the calculation [139, 140]. Goodness of linear fit is calculated using the ratio between sum of squares residual (SS_{res}) and sum of squares total (SS_{tot}) shown in equation 1. SS_{res} is the summation of square of the difference of each data point (y_i) from the best fit model (f_i), equation 2. This value is only mildly changed by an increase in the number of data points, particularly in a well fit model [139, 140]. SS_{tot}, the summation of the square of the difference of each data point (y_i) from the mean value of all data points ($\bar{y}$)(equation 3), however, is highly dependent on the range of values covered in the data set [139, 140]. The larger SS_{tot} becomes in relation to SS_{res} the better (higher) the linear fit. Thus, in a study comparing two groups or comparing results across studies, the number of points analyzed for each subject should be similar but, more importantly, the range of walking speeds should be comparable. It is important to note that this only applies to the range of speeds (i.e. difference from maximum to minimum), and that the absolute speeds do not have to match. The adjusted R^2, which normalizes the R^2 by the number of data points and the number of regressors, can be used to account for the difference in sample size but does not adjust for differences in the range of the data.

$$R^2 = 1 - \frac{SS_{res}}{SS_{tot}} \quad (1)$$

$$SS_{res} = \sum_i (y_i - f_i)^2 \quad (2)$$

$$SS_{tot} = \sum_i (y_i - \bar{y})^2 \quad (3)$$

The impact that the range of data has on the R^2 value may be favorable for some analyses using goodness of linear fit. An increase in walking speed range is reported to be a favorable health outcome [141] and would favorably increase the R^2 value. But, using different ranges to calculate R^2 for the purpose of quantifying and interpreting the coordination between stride length-cadence can mask important information. The sample data in figure 1 illustrates the stride length-cadence relationship for a subject walking at 3 (slow to fast) and 5 (very slow to very fast) self-selected walking speeds. Table 1 provides the number of data points, R^2, and adjusted R^2 values, along with the SS_{res} and SS_{tot} calculations. Visual inspection of the relationships in the 3-speed and 5-speed data sets shows an increase in the spread of the data points at the very slow end of the 5-speed range. This is confirmed by the 400% increase in SS_{res} with only a 97% increase in the number of data points. However, the R^2 is higher with the 5 speeds, even when adjusted for the number of data points due to the 1700% increase in SS_{tot}. So, while collecting 5 speeds may represent a broader spectrum of the subject's walking ability, the impact of the reduced coupling at speeds in the extreme ranges is lost due to the impact on SS_{tot}. While this illustration was a comparison within a single subject, the principle applies to other comparisons of R^2 values such as comparisons across groups with different walking speed ranges or comparisons pre/post intervention that could result in a change in walking speed range. Thus, when designing an experiment to evaluate the linear relationship between two variables that may vary in range, the aim of the study must consider how the range of data impacts the interpretation of the results.

Conclusion

The existing literature provides evidence of the need for new methods of assessment for lower-limb prosthesis users. While dual-task analysis has seen little successful use in evaluating prosthesis users, the methodology may still provide a useful framework for evaluating resource allocation in response to prosthetic use, particularly if dual-tasking is applied with novel analysis methods.

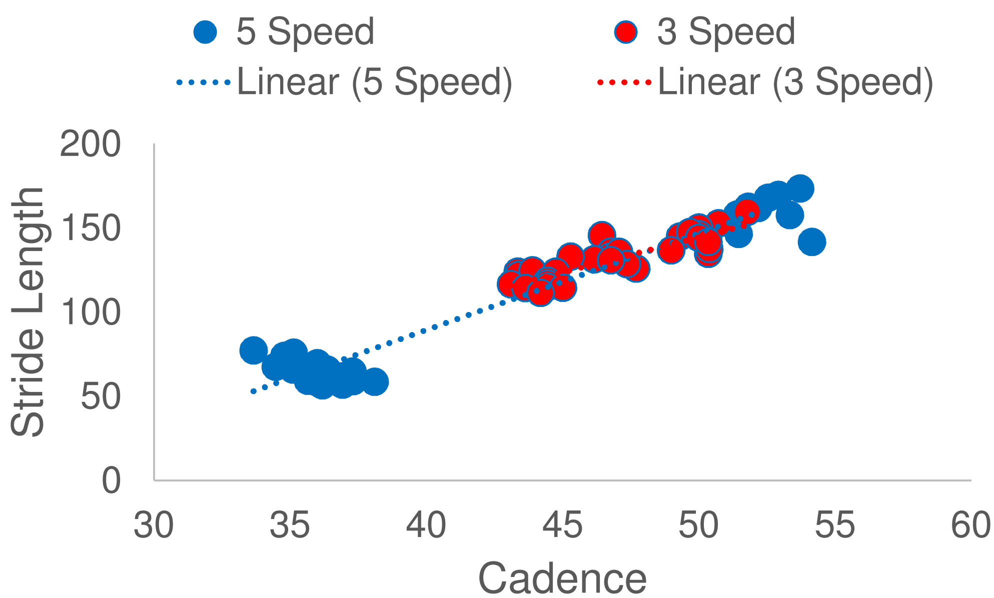

Fig. 1. Sample data of the stride length-cadence relationship comparing 3 and 5 self-selected speeds. Note the increased spread of the blue data in the left bottom corner, which is the very slow speed.

Table 1
Goodness of linear fit analysis of the stride length-cadence relationship for 3 and 5 self-selected speeds from figure 1.

	N	R^2	Adjusted R^2	SS_{res}	SS_{tot}
3 Speeds	34	0.764	0.749	1254	5327
5 Speeds	67	0.934	0.932	6293	96535

References

[1] Ku PX, Abu Osman NA, Wan Abas WA. Balance control in lower extremity amputees during quiet standing: A systematic review. Gait Posture. 2014; 39:672-82.

[2] Hunter SW, Batchelor F, Hill KD, Hill A-M, Mackintosh S, Payne M. Risk Factors for Falls in People With a Lower Limb Amputation: A Systematic Review. PMR. 2017; 9:170-80.

[3] Lamoth CJ, Ainsworth E, Polomski W, Houdijk H. Variability and stability analysis of walking of transfemoral amputees. Med Eng Phys. 2010; 32:1009-14.

[4] Miller WC, Speechley M, Deathe AB. The prevalence and risk factors of falling and fear of falling among lower extremity amputees. Arch Phys Med Rehabil. 2001; 82:1031-7.

[5] Smith DG, Michael JW, Bowker JH. Atlas of Amputations and Limb Deficiencies: Surgical, Prosthetic, and Rehabilitation Principles. 3rd ed: American Academy of Orthopaedic Surgeons; 2004.

[6] Klute GK, Berge JS, Orendurff MS, Williams RM, Czerniecki JM. Prosthetic intervention effects on activity of lower-extremity amputees. Arch Phys Med Rehabil. 2006; 87:717-22.

[7] Aldridge JM, Sturdy JT, Wilken JM. Stair ascent kinematics and kinetics with a powered lower leg system following transtibial amputation. Gait Posture. 2012; 36:291-5.

[8] Au S, Berniker M, Herr H. Powered ankle-foot prosthesis to assist level-ground and stair-descent gaits. Neural Netw. 2008; 21:654-66.

[9] Hafner BJ. Overview of outcome measures for the assessment of prosthetic foot and ankle componets. J Prosthet Orthot. 2006; 18:105.

[10] Klute GK, Kallfelz CF, Czerniecki J. Mechanical properties of prosthetic limbs:Adapting to the patient. J Rehabil Res Dev. 2001; 38:299-307.

[11] Hansen AH. Scientific Methods to determine functional performance of prosthetic ankle-foot systems. J Prosthet Orthot. 2005; 17:S23-S9.

[12] Segal AD, Orendurff MS, Czerniecki JM, Shofer JB, Klute GK. Local dynamic stability of amputees wearing a torsion adapter compared to a rigid adapter during straight-line and turning gait. J Biomech. 2010; 43:2798-803.

[13] Klodd E, Hansen AH, Fatone S, Edwards M. Effects of prosthetic foot forefoot flexibility on oxygen cost and subjective preference rankngs of unilateral transtibial prosthesis users. J Rehabil Res Dev. 2010; 47:543-52.

[14] McMulkin ML, Osebold WR, Mildes RD, Rosenquist RS. Comparison of three pediatric prosthetic feet during functional activities. J Prosthet Orthot. 2004; 16:78-84.

[15] Berge J, Czerniecki J, Klute GK. Efficacy of shock-absorbing versus rigid pylons for impact reduction in transtibial amputees based on laboratory, field, and outcome metrics. J Rehabil Res Dev. 2005; 42:795-808.

[16] Segal AD, Orendurff MS, Czerniecki JM, Shofer JB, Klute GK. Transtibial amputee joint rotation moments during straight-line walking and a common turning task with and without a torsion adapter. J Rehabil Res Dev. 2009; 46:375-83.

[17] Mohieldin A, Chidambaram A, Sabapathivinayagam R, Al Busairi W. Quantitative assessment of postural stability and balance between persons with lower limb amputation and normal subjects by using dynamic posturography. Maced J Med Sci. 2010; 3:138-43.

[18] Vanicek N, Strike S, McNaughton L, Polman R. Postural responses to dynamic perturbations in amputee fallers versus nonfallers: a comparative study with able-bodied subjects. Arch Phys Med Rehabil. 2009; 90:1018-25.

[19] Hermodsson Y, Ekdahl C, Persson B, Roxendal G. Standing balance in trans-tibial amputees following vascular disease or trauma: a comparative study with healthy subjects. Prosthet Orthot Int. 1994; 18:150-8.

[20] Vrieling AH, van Keeken HG, Schoppen T, Otten E, Hof AL, Halbertsma JP, et al. Balance control on a moving platform in unilateral lower limb amputees. Gait Posture. 2008; 28:222-8.

[21] Winter DA. Human balance and posture control during standing and walking. Gait Posture. 1995; 3:193-214.

[22] Buckley JG, O'Driscoll D, Bennett SJ. Postural sway and active balance performance in highly active lower-limb amputees. Am J Phys Med Rehabil. 2002; 81:13-20.

[23] Geurts AC, Mulder TW, Nienhuis B, Rijken RA. Postural reorganization following lower limb amputation. Possible motor and sensory determinants of recovery. Scand J Rehabil Med. 1992; 24:83-90.

[24] Isakov E, Mizrahi J, Ring H, Susak Z, Hakim N. Standing sway and weight-bearing distribution in people with below-knee amputations. Arch Phys Med Rehabil. 1992; 73:174-8.

[25] Gailey R, Allen K, Castles J, Kucharik J, Roeder M. Review of secondary physical conditions associated with lower-limb amputation and long-term prosthesis use. J Rehabil Res Dev. 2008; 45:15-29.

[26] Nolan L, Wit A, Dudzinski K, Lees A, Lake M, Wychowanski M. Adjustments in gait symmetry with walking speed in trans-femoral and trans-tibial amputees. Gait Posture. 2003; 17:142-51.

[27] Yeung LF, Leung AK, Zhang M, Lee WC. Long-distance walking effects on trans-tibial amputees compensatory gait patterns and implications on prosthetic designs and training. Gait Posture. 2012; 35:328-33.

[28] Highsmith MJ, Schulz BW, Hart-Hughes S, Latlief GA, Phillips SL. Differences in the spatiotemporal parameters of transtibial and transfemoral amputee gait. J Prosthet Orthot. 2010; 22:26-30

[29] Eberly VJ, Mulroy SJ, Gronley JK, Perry J, Yule WJ, Burnfield JM. Impact of a stance phase microprocessor-controlled knee prosthesis on level walking in lower functioning individuals with a transfemoral amputation. Prosthet Orthot Int. 2014; 38:447-55.

[30] Sawers A, Hafner BJ. Outcomes associated with the use of microprocessor-controlled prosthetic knees among individuals with unilateral transfemoral limb loss: A systematic review. JRRD. 2013; 50:273-314.

[31] Vanicek N, Strike S, McNaughton L, Polman R. Gait patterns in transtibial amputee fallers vs. non-fallers: Biomechanical differences during level walking. Gait Posture. 2009; 29:415-20.

[32] Schmalz T, Blumentritt S, Marx B. Biomechanical analysis of stair ambulation in lower limb amputees. Gait Posture. 2007; 25:267-78.

[33] Stark G. Perspectives on how and why feet are presecribed. J Prosthet Orthot. 2005; 17:S18-S22.

[34] Hafner BJ. Clinical prescription and use of prosthetic foot and ankle mechanisms: a review of the literature. J Prosthet Orthot. 2005; 17:S5-S11.

[35] Sawers AB, Hafner BJ. Outcomes associated with the use of microprocessor-controlled prosthetic knees among individuals with unilateral transfemoral limb loss: a systematic review. J Rehabil Res Dev. 2013; 50:273-314.

[36] Resnik L, Borgia M. Reliability of outcome measures for people with lower-limb amputations: distinguishing true change from statistical error. Phys Ther. 2011; 91:555-65.

[37] Klute GK, Glaister BC, Berge JS. Prosthetic liners for lower limb amputees: a review of the literature. Prosthet Orthot Int. 2010; 34:146-53.

[38] Gholizadeh H, Abu Osman NA, Eshraghi A, Ali S, Razak NA. Transtibial prosthesis suspension systems: systematic review of literature. Clin Biomech. 2014; 29:87-97.

[39] Hofstad C, Linde H, Limbeek J, Postema K. Prescription of prosthetic ankle-foot mechanisms after lower limb amputation. Cochrane Database Syst Rev. 2004; Epub: 10.1002/14651858.CD003978.pub2.

[40] van der Linde H, Hofstad CJ, Geurts AC, Postema K, Geertzen JH, van Limbeek J. A systematic literature review of the effect of different prosthetic components on human functioning with a lower-limb prosthesis. J Rehabil Res Dev. 2004; 41:555-70.

[41] Gholizadeh H, Abu Osman NA, Eshraghi A, Ali S. Transfemoral prosthesis suspension systems: a systematic review of the literature. Am J Phys Med Rehabil. 2014; 93:809-23.

[42] Hafner BJ, Sanders JE, Czerniecki J, Fergason J. Energy storage and return prostheses: does patient perception correlate with biomechanical analysis? Clin Biomech. 2002; 17:325-44.

[43] U. Raschke S, S. Orendurff M, Mattie JL, Kenyon DEA, Jones OY, Moe D, et al. Biomechanical characteristics, patient preference and activity level with different prosthetic feet: A randomized double blind trial with laboratory and community testing. J Biomech. 2015; 48:146-52.

[44] Miller LA, McCay JA. Summary and conclusion from the academy's sixth state-of-the-science conference on lower limb prosthetic outcome measures. SSC Proceedings. 2006; P2-P7.

[45] Geil MD. Recommendations for Research on Microprocessor Knees. J Prosthet Orthot. 2013; 25:P76-P9.

[46] Sawers A, Hahn ME, Kelly VE, Czerniecki J, Kartin D. Beyond componentry: How principles of motor learning can enhance locomotor rehabilitation of individuals with lower limb loss—A review. J Rehabil Res Dev. 2012; 49:1431-42.

[47] Woollacott M, Shumway-Cook A. Attention and the control of posture and gait: a review of an emerging area of research. Gait Posture. 2002; 16:1-14.

[48] Lacour M, Bernard-Demanze L, Dumitrescu M. Posture control, aging, and attention resources: Models and posture-analysis methods. Clin Neurophysiol. 2008; 38:411-21.

[49] Yogev-Seligmann G, Hausdorff JM, Giladi N. Do we always prioritize balance when walking? Towards an integrated model of task prioritization. Mov Disord. 2012; 27:765-70.

[50] Beurskens R, Steinberg F, Antoniewicz F, Wolff W, Granacher U. Neural correlates of dual-task walking: Effects of cognitive versus motor interference in young adults. Neural Plast. 2016; 2016:1-9.

[51] Bonnet CT, Baudry S. Active vision task and postural control in healthy, young adults: Synergy and probably not duality. Gait Posture. 2016; 48:57-63.

[52] Bisson EJ, Lajoie Y, Bilodeau M. The influence of age and surface compliance on changes in postural control and attention due to ankle neuromuscular fatigue. Exp Brain Res. 2014; 232:837-45.

[53] Bloem BR, Grimbergen YA, van Dijk JG, Munneke M. The "posture second" strategy: a review of wrong priorities in Parkinson's disease. J Neurol Sci. 2006; 248:196-204.

[54] Shumway-Cook A, Woollacott M, Kerns KA, Baldwin M. The effects of two types of cognitive tasks on postural stability in older adults with and without a history of falls. J Geront A. 1997; 52:232-40.

[55] Kelly VE, Eusterbrock AJ, Shumway-Cook A. Factors influencing dynamic prioritization during dual-task walking in healthy young adults. Gait Posture. 2013; 37:131-4.

[56] Tombu M, Jolicoeur P. A central capacity sharing model of dual-task performance. J Exp Psychol Hum Percept Perform. 2003; 29:3-18.

[57] Rosso AL, Cenciarini M, Sparto PJ, Loughlin PJ, Furman JM, Huppert TJ. Neuroimaging of an attention demanding dual-task during dynamic postural control. Gait Posture. 2017; 57:193-8.

[58] Pashler H. Graded capacity-sharing in dual-task interference? J Exp Psychol Hum Percept Perform. 1994; 20:330-42.

[59] Yogev-Seligmann G, Rotem-Galili Y, Dickstein R, Giladi N, Hausdorff JM. Effects of explicit prioritization on dual task walking in patients with Parkinson's disease. Gait Posture. 2012; 35:641-6.

[60] Plummer P, Eskes G. Measuring treatment effects on dual-task performance: a framework for research and clinical practice. Front Hum Neurosci. 2015; 9:225.

[61] Wrightson JG, Twomey R, Ross EZ, Smeeton NJ. The effect of transcranial direct current stimulation on task processing and prioritisation during dual-task gait. Exp Brain Res. 2015; 233:1575-83.

[62] Swanenburg J, de Bruin ED, Favero K, Uebelhart D, Mulder T. The reliability of postural balance measures in single and dual tasking in elderly fallers and non-fallers. BMC Musculoskelet Disord. 2008; 9:162.

[63] Plummer-D'Amato P, Altmann LJP, Saracino D, Fox E, Behrman AL, Marsiske M. Interactions between cognitive tasks and gait after stroke: A dual task study. Gait Posture. 2008; 27:683-8.

[64] McIsaac TL, Lamberg EM, Muratori LM. Building a framework for a dual task taxonomy. Biomed Res Int. 2015; 2015:1-10.

[65] Wrightson JG, Ross EZ, Smeeton NJ. The Effect of Cognitive-Task Type and Walking Speed on Dual-Task Gait in Healthy Adults. Motor Control. 2016; 20:109-21.

[66] Fok P, Farrell M, McMeeken J. Prioritizing gait in dual-task conditions in people with Parkinson's. Hum Mov Sci. 2010; 29:831-42.

[67] Hamacher D, Brennicke M, Behrendt T, Alt P, Torpel A, Schega L. Motor-cognitive dual-tasking under hypoxia. Exp Brain Res. 2017; Epub: Jul 18; 10.1007/s00221-017-5036-y.

[68] Hollman JH, Youdas JW, Lanzino DJ. Gender differences in dual task gait performance in older adults. Am J Mens Health. 2011; 5:11-7.

[69] Beauchet O, Dubost V, Aminian K, Gonthier R, Kressig RW. Dual-task-related gait changes in the elderly: does the type of cognitive task matter? J Mot Behav. 2005; 37:259-64.

[70] Williams RM, Turner AP, Orendurff M, Segal AD, Klute GK, Pecoraro J, et al. Does having a computerized prosthetic knee influence cognitive performance during amputee walking? Arch Phys Med Rehabil. 2006; 87:989-94.

[71] Olivier I, Cuisinier R, Vaugoyeau M, Nougier V, Assaiante C. Age-related differences in cognitive and postural dual-task performance. Gait Posture. 2010; 32:494-9.

[72] Geurts AC, Mulder TW, Nienhuis B, Rijken RA. Dual-task assessment of reorganization of postural control in persons with lower limb amputation. Arch Phys Med Rehabil. 1991; 72:1059-64.

[73] Lindenberger U, Marsiske M, Baltes PB. Memorizing while walking: increase in dual-task costs from young adulthood to old age. Psychol Aging. 2000; 15:417.

[74] Reissland J, Manzey D. Serial or overlapping processing in multitasking as individual preference: Effects of stimulus preview on task switching and concurrent dual-task performance. Acta Psychol 2016; 168:27-40.

[75] Yang YR, Chen YC, Lee CS, Cheng SJ, Wang RY. Dual-task-related gait changes in individuals with stroke. Gait Posture. 2007; 25:185-90.

[76] Nordin E, Moe-Nilssen R, Ramnemark A, Lundin-Olsson L. Changes in step-width during dual-task walking predicts falls. Gait Posture. 2010; 32:92-7.

[77] Lamberg EM, Muratori LM. Cell phones change the way we walk. Gait Posture. 2012; 35:688-90.

[78] Huang CY, Lin LL, Hwang IS. Age-Related Differences in Reorganization of Functional Connectivity for a Dual Task with Increasing Postural Destabilization. Front Aging Neurosci. 2017; 9:96.

[79] Huang HJ, Mercer VS. Dual-task methodology: applications in studies of cognitive and motor performance in adults and children. Pediatr Phys Ther. 2001; 13:133-40.

[80] Bernard-Demanze L, Dumitrescu M, Jimeno P, Borel L, Lacour M. Age-related changes in posture control are differentially affected by postural and cognitive task complexity. Curr Aging Sci. 2009; 2:139-49.

[81] Salavati M, Mazaheri M, Negahban H, Ebrahimi I, Jafari AH, Kazemnejad A, et al. Effect of dual-tasking on postural control in subjects with nonspecific low back pain. Spine. 2009; 34:1415-21.

[82] Morgan SJ, Hafner BJ, Kelly VE. Dual-task walking over a compliant foam surface: A comparison of people with transfemoral amputation and controls. Gait Posture. 2017; 58:41-5.

[83] Gimmon Y, Jacob G, Lenoble-Hoskovec C, Büla C, Melzer I. Relative and absolute reliability of the clinical version of the Narrow Path Walking Test (NPWT) under single and dual task conditions. Arch Gerontol Geriatr. 2013; 57:92-9.

[84] Kelly VE, Schrager MA, Price R, Ferrucci L, Shumway-Cook A. Age-associated effects of a concurrent cognitive task on gait speed and stability during narrow-base walking. J Geront A. 2008; 63:1329-34.

[85] Amboni M, Barone P, Hausdorff JM. Cognitive contributions to gait and falls: evidence and implications. Mov Disord. 2013; 28:1520-33.

[86] Lord S, Howe T, Greenland J, Simpson L, Rochester L. Gait variability in older adults: A structured review of testing protocol and clinimetric properties. Gait Posture. 2011; 34:443-50.

[87] Springer S, Giladi N, Peretz C, Yogev G, Simon ES, Hausdorff JM. Dual-tasking effects on gait variability: the role of aging, falls, and executive function. Mov Disord. 2006; 21:950-7.

[88] Ijmker T, Lamoth CJC. Gait and cognition: The relationship between gait stability and variability with executive function in persons with and without dementia. Gait Posture. 2012; 35:126-30.

[89] Kressig RW, Herrmann FR, Grandjean R, Michel JP, Beauchet O. Gait variability while dual-tasking: fall predictor in older inpatients? Aging Clin Exp Res. 2008; 20:123-30.

[90] Sample RB, Jackson K, Kinney AL, Diestelkamp WS, Reinert SS, Bigelow KE. Manual and Cognitive Dual Tasks Contribute to Fall-Risk Differentiation in Posturography Measures. J Appl Biomech. 2016; 32:541-7.

[91] Fujita H, Kasubuchi K, Wakata S, Hiyamizu M, Morioka S. Role of the Frontal Cortex in Standing Postural Sway Tasks While Dual-Tasking: A Functional Near-Infrared Spectroscopy Study Examining Working Memory Capacity. Biomed Res Int. 2016; Epub: Feb 3; 10.1155/2016/7053867.

[92] Collins JJ, De Luca CJ, Burrows A, Lipsitz LA. Age-related changes in open-loop and closed-loop postural control mechanisms. Exp Brain Res. 1995; 104:480-92.

[93] Ghulyan V, Paolino M, Lopez C, Dumitrescu M, Lacour M. A new translational platform for evaluating aging or pathology-related postural disorders. Acta Otolaryngol. 2005; 125:607-17.

[94] Geurts AC, Mulder TH. Attention demands in balance recovery following lower limb amputation. J Mot Behav. 1994; 26:162-70.

[95] Morgan SJ, Hafner BJ, Kelly VE. The effects of a concurrent task on walking in persons with transfemoral amputation compared to persons without limb loss. Prosthet Orthot Int. 2016; 40:490-6.

[96] Aggashyan RV, Gurfinkel VS, Mamasakhlisov GV, Elner AM. Changes in spectral and correlation characteristics of human stabilograms at muscle afferentation disturbance. Agressologie. 1973; 14:5-9.

[97] Bensel CK, Dzendolet E. Power spectral density analysis of the standing sway of males. Atten Percept Psychophys. 1968; 4:285-8.

[98] Diener HC, Dichgans J, Bruzek W, Selinka H. Stabilization of human posture during induced oscillations of the body. Exp Brain Res. 1982; 45:126-32.

[99] Diener HC, Dichgans J, Guschlbauer B, Mau H. The significance of proprioception on postural stabilization as assessed by ischemia. Brain Res. 1984; 296:103-9.

[100] Collins JJ, De Luca CJ. Open-loop and closed-loop control of posture: a random-walk analysis of center-of-pressure trajectories. Exp Brain Res. 1993; 95:308-18.

[101] Collins JJ, De Luca CJ. Random walking during quiet standing. Phys Rev Lett. 1994; 73:764-7.

[102] Singh NK, Snoussi H, Hewson D, Duchêne J. Wavelet transform analysis of the power spectrum of centre of pressure signals to detect the critical point interval of postural control. Biomedical Engineering Systems and Technologies: Springer; 2010. p. 235-44.

[103] Soames RW, Atha J. The spectral characteristics of postural sway behaviour. Eur J Appl Physiol Occup Physiol. 1982; 49:169-77.

[104] Patla AE, Ishac MG, Winter DA. Anticipatory control of center of mass and joint stability during voluntary arm movement from a standing posture: interplay between active and passive control. Exp Brain Res. 2002; 143:318-27.

[105] Winter DA, Patla AE, Ishac M, Gage WH. Motor mechanisms of balance during quiet standing. J Electromyogr Kinesiol. 2003; 13:49-56.

[106] Morasso PG, Schieppati M. Can muscle stiffness alone stabilize upright standing? J Neurophysiol. 1999; 82:1622-6.

[107] Morasso PG, Sanguineti V. Ankle muscle stiffness alone cannot stabilize balance during quiet standing. J Neurophysiol. 2002; 88:2157-62.

[108] Peterka R. Sensorimotor integration in human postural control. J Neurophys. 2002; 88:1097-118.

[109] Scott DE, Dzendolet E. Quantification of sway in standing humans. Agressologie. 1972; 13:Suppl B:35-4.

[110] Oppenheim U, Kohen-Raz R, Alex D, Kohen-Raz A, Azarya M. Postural characteristics of diabetic neuropathy. Diabetes care. 1999; 22:328-32.

[111] Mauritz KH, Dichgans J, Hufschmidt A. Quantitative analysis of stance in late cortical cerebellar atrophy of the anterior lobe and other forms of cerebellar ataxia. Brain. 1979; 102:461-82.

[112] Diener HC, Dichgans J, Bacher M, Gompf B. Quantification of postural sway in normals and patients with cerebellar diseases. Electroencephalogr Clin Neurophysiol. 1984; 57:134-42.

[113] Thurner S, Mittermaier C, Hanel R, Ehrenberger K. Scaling-violation phenomena and fractality in the human posture control systems. Phys Rev E Stat Phys Plasmas Fluids Relat Interdiscip Topics. 2000; 62:4018-24.

[114] Kirchner M, Schubert P, Schmidtbleicher D, Haas C. Evaluation of the temporal structure of postural sway fluctuations based on a comprehensive set of analysis tools. Phys A. 2012; 391:4692-703.

[115] Chagdes JR, Rietdyk S, Haddad JM, Zelaznik HN, Raman A, Rhea CK, et al. Multiple timescales in postural dynamics associated with vision and a secondary task are revealed by wavelet analysis. Exp Brain Res. 2009; 197:297-310.

[116] Mauritz KH, Dietz V. Characteristics of postural instability induced by ischemic blocking of leg afferents. Exp Brain Res. 1980; 38:117-9.

[117] Roeing KL, Wajda DA, Sosnoff JJ. Time dependent structure of postural sway in individuals with multiple sclerosis. Gait Posture. 2016; 48:19-23.

[118] Martinez-Ramirez A, Lecumberri P, Gomez M, Izquierdo M. Wavelet analysis based on time-frequency information discriminate chronic ankle instability. Clin Biomech. 2010; 25:256-64.

[119] Martinez-Ramirez A, Lecumberri P, Gomez M, Rodriguez-Manas L, Garcia FJ, Izquierdo M. Frailty assessment based on wavelet analysis during quiet standing balance test. J Biomech. 2011; 44:2213-20.

[120] Torrence C, Compo GP. A practical guide to wavelet analysis. Bull Amer Meteor. 1998; 79:61-78.

[121] Misiti M, Misiti Y, Oppenheim G, Poggi JM. Wavelet Toolbox for use with Matlab. Wavelet Toolbox User's Guide: The MathWorks, Inc.; 1996.

[122] In F, Kim S. Introduction to Wavelet Theory in Finance : A Wavelet Multiscale Approach. Singapore, UNITED STATES: World Scientific Publishing Company; 2012.

[123] Peters M. Footedness: asymmetries in foot preference and skill and neuropsychological assessment of foot movement. Psychol Bull. 1988; 103:179-92.

[124] Elias LJ, Bryden MP. Footedness is a better predictor of language lateralisation than handedness. Laterality. 1998; 3:41-51.

[125] Elias LJ, Bryden MP, Bulman-Fleming MB. Footedness is a better predictor than is handedness of emotional lateralization. Neuropsychologia. 1998; 36:37-43.

[126] Taylor MJ, Strike SC, Dabnichki P. Turning bias and lateral dominance in a sample of able-bodied and amputee participants. Laterality. 2006; 12:50-63.

[127] Grouios G. Footedness as a potential factor that contributes to the causation of corn and callus formation in lower extremities of physically active individuals. The Foot. 2005; 15:154-62.

[128] Sadeghi H, Allard P, Prince F, Labelle H. Symmetry and limb dominance in able-bodied gait: a review. Gait Posture. 2000; 12:34-45.

[129] Dessery Y, Barbier F, Gillet C, Corbeil P. Does lower limb preference influence gait initation? Gait Posture. 2011; 33:550-5.

[130] Yazgan MY, Leckman JF, Wexler BE. A direct observational measure of whole body turning bias. Cortex. 1996; 32:173-6.

[131] Kalaycioglu C, Kara C, Atbasoglu C, Nalcaci E. Aspects of foot preference: differential relationships of skilled and unskilled foot movements with motor asymmetry. Laterality. 2008; 13:124-42.

[132] Egerton T, Danoudis M, Huxham F, Iansek R. Central gait control mechanisms and the stride length - cadence relationship. Gait Posture. 2011; 34:178-82.

[133] Zijlstra W, Rutgers AWF, Hof AL, Van Weerden TW. Voluntary and involuntary adaptation of walking to temporal and spatial constraints. Gait Posture. 1995; 3:13-8.

[134] Du Chatinier K, Molen NH, Rozendal RH. Step length, step frequency and temporal factors of the stride in normal human walking. Anatomy. 1970; 73:214-27.

[135] Andriacchi TP, Ogle JA, Galante JO. Walking speed as a basis for normal and abnormal gait measurments. J Biomech. 1977; 10:261-8.

[136] Morris M, Iansek R, Matyas T, Summers J. Abnormalities in stride length-cadence relation in parkinsonian gait. Mov Disord. 1998; 13:61-9.

[137] Grieve DW, Gear RJ. The relationships between length of stride, step frequency, time of swing and speed of walking for children and adults. Ergonomics. 1966; 5:379-99.

[138] Murray MP, Kory RC, Clarkson BH, Sepic SB. Comparison of free and fast speed walking patterns of normal men. PMR. 1966; 45:8-24.

[139] Montgomery DC, Peck EA, Vining GG. Introduction to linear regression analysis: John Wiley & Sons; 2012.

[140] Seltman HJ. Experimental design and analysis. Pittsburgh: Carnegie Mellon University; 2012.

[141] Menant JC, Schoene D, Sarofim M, Lord SR. Single and dual task tests of gait speed are equivalent in the prediction of falls in older people: a systematic review and meta-analysis. Ageing Res Rev. 2014; 16:83-104.

CHAPTER 3

LOWER LIMB PREFERENCE ON GOAL-ORIENTED TASKS IN UNILATERAL PROSTHESIS USERS

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Abstract

The aim of this study was to determine lower limb preference in 31 prosthesis users and 19 able-bodied controls on 11 goal-oriented tasks in free-standing and supported conditions. The action leg used in 6 or more tasks was considered the preferred leg. We hypothesized that the prosthetic leg in amputees would be used as the preferred leg as often as the dominant leg in controls. For prosthesis users in the free-standing condition, 65% used the prosthetic leg as the preferred leg. This was significantly different ($p<0.003$) from able-bodied controls, where 100% used the dominant leg as the preferred leg. This discrepancy became even more pronounced in the supported condition and was overall more prevalent among those who used prosthesis for more than 10 years. These findings may have implications for therapy and gait training.

Introduction

Just as hand dominance enables us to predict how people write or throw a ball, foot dominance determines how people perform tasks with their lower limbs [1]. Foot dominance is considered an innate preference stemming from cerebral lateralization, as it has been linked with language and emotional lateralization [2, 3]. Peters [1] provided the most commonly used definition for foot dominance: “the foot that is used to manipulate an object or to lead out, as in jumping, is deemed here as the preferred foot. The foot that is used to support the activities of the preferred foot by lending postural and stabilizing support is defined as the non-preferred foot”. The preferred or dominant foot is consistent across most goal-oriented lower limb tasks in healthy people [1].

Acquired unilateral lower-limb amputation provides a unique opportunity for studying changes in lower limb preference. As opposed to acting on their innate preference, amputees fitted with a prosthetic device must deal with the quandary of compromising between stability and performance. Several scenarios are possible when considering how lower limb preference may be altered in prosthesis users. One possibility is that prosthesis users would resort to the strategy that presumably provides the most stable state when performing lower limb tasks. That is, they may opt to rely on their intact limb for stability. This corresponds with standard stair training where many prosthesis users receive advice to use the intact side as the primary supporting limb [4]. Another scenario is that cerebral dominance may still prevail and, tied with lifelong habit, could influence prosthesis users to maintain their previous strategy despite limb loss. The selection of strategy by prosthesis users may further be influenced by motivation for goal achievement, speed and accuracy required for the task, the residual limb length, and time since amputation. For example, without active ankle or knee motion, it may be difficult for prosthesis users to adequately manipulate an object, thus encouraging the use of the intact limb as the preferred limb regardless of previous dominance. The latter assumption may particularly hold if stability is not compromised, such as when support is available.

Thus, the choice of strategy may depend on the interplay between settings in which the task is to be performed, desire to complete the task, fear of falling, as well as prosthesis fit and experience. It is clear, therefore, that the common assumptions of action and stabilizing leg roles in able-bodied individuals may not translate to lower-limb prosthesis users.

The purpose of this study was to examine how amputation alters lower limb preference in prosthesis users and to explore some potentially contributing factors. We hypothesized that prosthesis users will use the intact leg for stability and the prosthetic leg for performance across different goal-oriented tasks with the same consistency as able-bodied subjects use the non-dominant and dominant legs, respectively. We specifically tested whether the prosthetic leg in amputees is used as the preferred leg as often as the dominant leg in controls. We also explored how upper limb support affects performance strategy under the assumption that lower limb preference will become more apparent from the free-standing to supported condition. The potential role of residual limb length, side, and time since amputation was examined in secondary analysis. Along with task performance, limb preference has been related to several aspects of gait [5, 6], including turning [7], gait initiation [8], and stair climbing [9]. Therefore, these results are expected to improve understanding of motor control strategies utilized by prosthesis users and may have implications for therapy.

Methods

Participants

We recruited unilateral above- and below-knee prosthesis users from 5 prosthetic clinics run by our institution throughout Mississippi and Louisiana. The inclusion criteria were: (1) acquired lower limb loss, (2) use of a prosthesis for over a year, (3) age 18 – 80 years, (4) comfortable socket fit, (5) healthy residual limb, (6) healthy contralateral limb, (7) no use of assistive device for everyday activities, (8) no known balance, neurological, or other health problems that limit daily activities, (9) verification by certified prosthetist that prosthesis user was fit to attempt experimental tasks. A convenience sample of age-matched able-body control subjects was also recruited with the following inclusion criteria: (1) age 18 – 80 years, (2) no use of an assistive device, (3) no balance, neurological, orthopedic, or general health problems that limit daily activities.

The study sample included 19 able-bodied controls (9 men; mean age 42±13.5 years, 18 right-handed) and 31 prosthesis users (20 men; mean age 49±14.2 years; 27 right-handed; 20 below-knee amputees and 11 above-knee amputees). The average time since amputation was 13.2 years (range 1.9 to 43 years). The amputation was due to trauma (n=23), vascular disease (n=4), and other causes (n=4). The subjects were rated K3 (n=29) or K4 (n=2) on the Medicare scale. Both prosthesis users and controls were of average stature. The study protocol was approved by the institutional review board for human research and all subjects signed an informed consent form. Prosthesis users wore their primary prosthesis and all subjects wore their own shoes during testing. Data for all subjects were collected by the same researcher at five prosthetic clinics and an in-patient rehabilitation and research facility.

Protocol

We developed the Assessment of Leg Preference in Amputees (Table 1) for this study, which was done under two conditions: free-standing in an open area (condition 1) and standing with hands on parallel bars (supported, condition 2). The tasks were selected from the literature on lower limb laterality [1-3, 9, 10] and input from a certified prosthetist and physical therapist. The tasks were goal-oriented and encompassed typical motions performed with the lower limbs. Each subject performed 11 tasks in each condition: five tasks were identical in both conditions and the remaining six were each selected from six pairs of tasks. Each task within a pair was randomly assigned to the free-standing or supported condition. Paired tasks were used so that virtually the same motion would be required under each condition, but the slight differences in task would deter the subject from recalling the previous action. Paired tasks were of similar difficulty and each one was cued with different objects.

A start line and midline with marks at 15, 25, and 35 cm were taped on the floor to ensure tasks were presented in the same manner. The subject assumed a natural stance with feet equally spaced from the midline. The lower limbs were video recorded so the researcher could direct attention to the subject and later analyze data.

Prosthesis users were tested in free-standing and supported conditions in random order. Eleven able-body subjects were tested in both the free-standing and supported conditions and demonstrated high consistency in performance between the two conditions (98% agreement for task pairs). Thus, an additional eight subjects performed the free-standing condition only. The 11 tasks in each condition were presented in a random order. A seated break was given between the two conditions when demographic information was collected.

We were concerned that knowledge of the purpose of the experiment prior to testing may influence the subjects' performance. Therefore, subjects were told that the purpose of the study was to examine their ability to, rather than how they perform each task. Subjects were instructed to perform each task in the most comfortable way as if encountered in their daily life. No suggestion was given as to which leg to use. If asked, the researcher replied that the choice of leg was not being examined.

Upon completion of the tasks, all subjects filled out the Waterloo Handedness Questionnaire-Revised (WHQ-R) [2] to determine hand dominance. One month later, to avoid bias from completing the WHQ-R and task performance, they were mailed the Waterloo Footedness Questionnaire-Revised (WFQ-R) [2] to determine perceived leg dominance.

Data processing

Due to presentation of multiple tasks, the language used by Peters [1] for leg categorization has been slightly altered. When analyzing specific tasks, the leg used to perform the task is referred to as the *action leg*. For example, the foot that makes contact with the cueing object, such as in kicking a ball, or the leg that leads out to step over an object is considered the action leg for that task. The action leg also served to appreciate the consistency in tasks performance.

The predominant choice of action leg across the presented tasks was used to define the preferred leg for each condition. That is, the leg used 6 or more times as the action leg out of the 11 tasks was considered the preferred leg for the free-standing or supported condition, respectively. Thus, the primary outcome variables were the action leg for each task and the preferred leg for each condition.

Table 1

Description of the tasks and administration protocol for the Assessment of Leg Preference in Amputees developed for this study.

#	Task	Cueing Object	Instruction	Distance from Subject
1	Rolling Kick	20cm diameter ball and two cones	**Kick** a ball rolled down a ramp at either of two equally spaced cones	Start:175cm Contact: Variable Cones: 175cm away, 60cm wide
2	Elevator Door	Line on floor	**Stick out** foot as if trying to stop an elevator door from closing	25cm
3	Open Lid	60x35x145cm wooden box with hinged lid (2cm lip)	**Lift** the lid with a foot approximately 15cm (limited by a cord connected to the lid) and then close it	25cm
4	Stationary Kick	20cm diameter ball and two cones	**Kick** a stationary ball at either of two equally spaced cones	Ball: 25cm Cones:175cm away, 60cm apart
5	Moving Target	15cm diameter disk affixed to retracting tape	**Step on** disk before it crosses the start position line	Start:175cm Contact: Variable
6a/b	Garbage Can/ Pump	Foot pedal garbage can/ Foot operated air pump	**Press and release** garbage can pedal/Operate **pump** three times	15cm
7a/b	Match/Bug	Fake match/Fake bug	**Stomp** to extinguish fire/exterminate bug	25cm
8a/b	Push Box/Ball	30x18x9cm box/ 20cm diameter ball	**Push** box/ball to either side with foot	Center at 25cm
9a/b	Step Box/Animal	30x18x9cm box/ medium sized stuffed animal	**Step over** the box/animal with one foot	Center at 25cm
10a/b	Smooth/Wipe	100x50cm wrinkled towel/water in 40x20cm area and small cloth	**Swipe** foot to straighten towel/clean spill	Center at 25cm
11a/b	Pencil/CD	18cm pencil/ CD case	**Reach** out and **pull** pencil/CD back	35cm

The video recording was reviewed to assign the action leg to right or left side for each task. The action leg was subsequently translated into 1) prosthetic or intact leg for the amputee population and 2) dominant or non-dominant leg for the control population. The preferred leg was then assigned to each condition and identically translated.

Hand dominance was determined based on WHQ-R. Leg dominance was defined in two ways. First, the dominant leg was considered the one used to kick the rolling ball. Secondly, the perceived leg dominance was defined based on the response to the first question of the WFQ-R questionnaire (kicking a ball). The agreement of perceived leg dominance with the action leg used in the rolling and stationary kick tasks and the preferred leg across all tasks was analyzed for all controls and 26 (84%) prosthesis users who returned the questionnaire. Those who indicated no leg preference on WFQ-R were excluded from the latter analysis (1 control, 2 prosthesis users).

Statistical analysis

Frequency histograms were used to describe the distribution of action leg across multiple tasks for each condition and subject group. Fisher's exact test was used to test the null hypothesis that the prosthetic leg in amputees was used as the preferred leg as often as the dominant leg in controls. The change in preferred leg from free-standing to supported standing was also examined with the Fisher's exact test in each group. The same test was also used in secondary analyses to explore whether the choice of preferred leg (prosthetic vs. intact) differed between below- and above-knee amputees, side of amputation, or with time since amputation (1-10 years, >11 years).

Results

Action leg in free-standing condition

When control subjects performed tasks in the free-standing condition, the action leg largely corresponded to the dominant hand across different tasks (mean 82%, range 74-84%). When leg dominance was determined by the ball kicking task, the action leg in the remaining tasks almost perfectly matched the kicking (dominant) leg (mean 97%, range 89-100%). In the prosthesis users, however, the choice of action leg was less consistent. When pooled across all prosthesis users, the action leg matched the prosthetic leg in 56% of tasks, on average. Therefore, the prosthetic leg was used less often as the action leg than the dominant (kicking) leg in the controls. The analysis across tasks in the prosthesis users indicated that the action leg matched the prosthetic side least often in the bug/match stomping (35%) and garbage can/pumping (39%) tasks, most often in the elevator door stopping (81%), lid opening (81%), and box/ball pushing (74%) tasks, and in about 50% of other tasks. Figure 1 shows the correspondence of action leg with the dominant (kicking) leg in the controls and with the prosthetic leg in prosthesis users across different tasks in the free-standing condition.

In general, the controls were likely to choose the same action leg for all tasks. Prosthesis users, on the other hand, were more likely to switch their action leg between tasks. The greater consistency in controls than prosthesis users is summarized in Figure 2. The number of switching instances across different tasks was no more than 2 in the control group whereas it ranged from 1-9 in the prosthesis users.

Comparision of action leg between free and supported condition

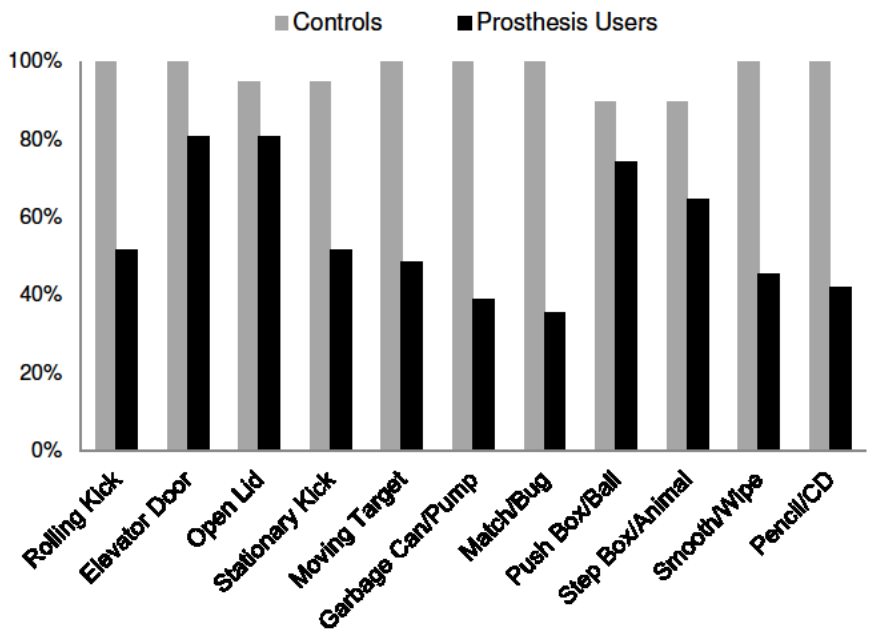

Fig. 1. Performance on 11 lower limb tasks. (White) Percent of controls who used their dominant (kicking) leg as the action leg in each task. (Black) Percent of prosthesis users who used their prosthetic leg as the action leg in each task.

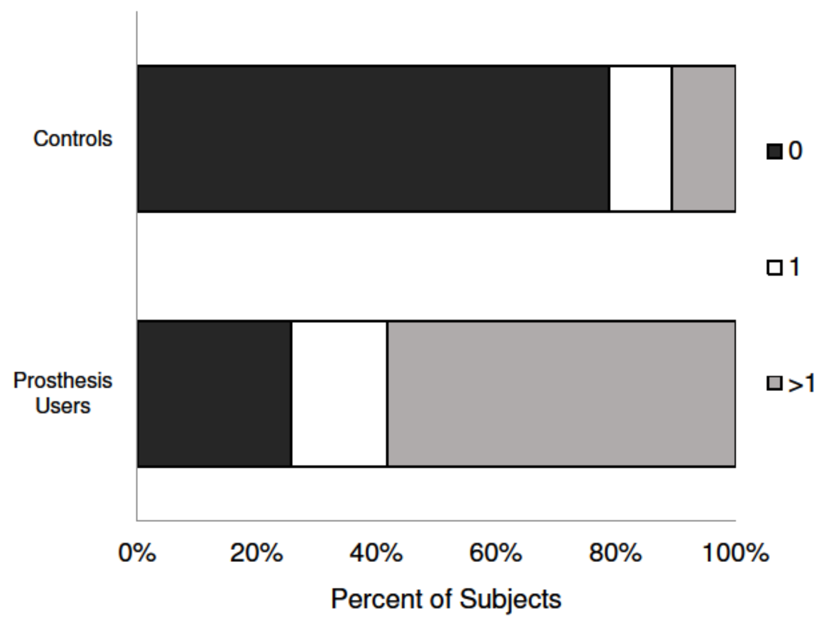

Fig. 2. Percent of subjects who performed tasks with a different leg than the one used to kick the rolling ball in the free-standing condition (0- never, 1- once, >1- more than once). Note higher rate of switching among the prosthesis users.

Matched tasks were used to compare performance between two conditions. Controls performed rather consistently across the two conditions. Specifically, 9 of 11 (82%) control subjects who were examined in both conditions used the same action leg in all matched tasks. The remaining two differed in only one task.

The prosthesis users, however, demonstrated marked differences between the free and supported standing conditions. Only 26% (8/31) of the prosthesis users used the same action leg in all matched tasks between the two conditions. Among the remaining 23 prosthesis users, the average number of tasks performed differently was 4±2. Seventeen of 23 (74%) who switched action legs in the matched tasks went mainly from the prosthetic leg during free-standing to the intact leg in supported standing. Twelve of these 17 (71%) consistently switched in this manner.

Preferred leg in free-standing condition

The preferred leg matched the kicking leg for all 19 control subjects and 27 (87%) of prosthesis users in the free-standing condition. In the prosthesis users however, the preferred leg matched the prosthetic side in 20 of 31 (65%). The prosthetic leg was used significantly less often as the preferred leg than the dominant (kicking) leg in the controls (Fisher's exact test $p<0.003$).

Comparision of preferred leg between free and supported condition

As with the free-standing condition, the preferred leg was the the same as the kicking leg in all control subjects for the supported condition, but only in 26 (84%) of the prosthesis users. Nine prosthesis users switched the preferred leg between the two conditions. In all 9, the preferred leg changed from prosthetic side during the free-standing condition to the intact side during the supported condition. This change in the prosthetic users was significant (Fisher's exact test, $p<0.04$).

Relationship to characteristics of amputation

Table 2 shows the distribution of the preferred leg by different amputation characteristics. Having a prosthesis for more than 10 years was significantly associated with more frequent use of the intact leg as the preferred leg under both conditions (Fisher's exact test, $p=0.02$ free, $p=0.008$ supported). No significant difference was found for the residual limb length ($0.13 < p < 0.70$) or side with respect to hand dominance ($0.15 < p < 0.70$).

Table 2
The distribution of preferred leg in relation to time, level, and side of amputation (significant difference indicated in bold).

	Free-Standing Condition			Supported Condition		
	Prosthetic Leg	Intact Leg	p-Value	Prosthetic Leg	Intact Leg	p-value
Time since Amputation						
>10 yrs.	5	8		1	12	
≤10 yrs.	15	3	**0.02**	10	8	**0.008**
Level of Amputation						
Above-Knee	5	6		3	8	
Below-Knee	15	5	0.13	8	12	0.70
Side of Amputation						
Hand Dominant	13	4		7	10	
Hand Non-Dominant	7	7	0.15	4	10	0.70

Table 3
The rate of agreement between the endorsed kicking leg on WFQ-R and the actual leg used in two kicking tasks and the overall preferred leg (controls: free-standing n=18, supported n=10; prosthesis users: n=24 for both).

	Stationary Kick		Rolling Kick		Preferred Leg	
	Free-Standing	Supported	Free-Standing	Supported	Free-Standing	Supported
Controls	94%	100%	100%	100%	100%	100%
Prosthesis Users	71%	62%	62%	75%	67%	67%

WFQ-R

The agreement between the endorsed kicking leg on WFQ-R and the actual kicking and preferred leg was nearly perfect in the controls, but was inconsistent in 25-38% of prosthesis users (Table 3).

Discussion

The main result of this study is that prosthesis users do not consistently use their prosthetic leg when performing different goal-oriented lower limb tasks. Thus, we refute our main hypothesis that the prosthesis users choose the intact leg for stability and the prosthetic leg for performance with the same consistency as able-bodied persons use the non-dominant and dominant legs, respectively. These findings were reaffirmed by the observation that when arm support was provided, prosthesis users increased their preference toward completing tasks with intact leg while standing on the prosthetic leg, but no change was observed in controls. This strategy was more prevalent in more experienced prosthesis users. Finally, the discrepancy between the perceived and actual leg preference was evident in 25-38% of prosthesis users and none of the controls.

Our results contradict some common assumptions and provide important insight into motor behavior of prosthesis users. In the able-bodied population, lower limb laterality is used to appreciate how individuals maneuver through the world, including normal gait [5, 6, 8], turning [7] and stair stepping [9]. Our findings of less prominent leg preference in the prosthesis users may provide basis for some unexpected motor behavior during different activities. Taylor et al. reported a preference toward turning to the left among right-handed controls, which was not found in right-handed below-knee prosthesis users. Although prosthesis users showed a trend towards turning towards the prosthetic side, no significant factor was identified to predict turning bias [7]. This echoes our results because the side of amputation with respect to hand dominance was not associated with goal-oriented leg preference indicating poor predictability of leg preference based on side of amputation.

We found that only time since amputation was associated with leg preference. More experienced users were more likely to rely on their intact side as their preferred leg. After years of use, they may have become more comfortable or trusting of their prosthesis. As such, they are able to utilize their prosthesis for balance and benefit from active motion of the intact side for task performance. While this change may result from years of practice, neural changes should not be overlooked. There have been only a few studies examining neural adaptation after lower-limb amputation. While they suggest motor reorganization occurs at the cortical level, it is unclear how it translates into motor action in the lower limbs [11].

The overall tendency of prosthesis users to use the intact side as the action or preferred leg became more apparent during supported standing, when balance is not compromised. The prosthesis users can then safely stand on the prosthetic leg and more precisely manipulate the object with the intact leg. This strategy may have been selected because of mechanical limitations of the prosthetic device, which may hamper successful completion of tasks.

Conclusion

Our findings have several implications for clinical practice and research. Whereas current rehabilitation practice is focused on retraining level walking and stair climbing in prosthesis users[4], we suggest a broader inclusion of activities to train the prosthetic leg in both stability and performance tasks. This would allow a prosthesis user to experience a variety of daily tasks and facilitate the development of individual strategy. Inconsistency between the questionnaire response and actual performance reinforces the idea that prosthesis users lack strategy for goal-oriented tasks. This may delay reaction time and pose a risk in unfamiliar settings. Early acquisition of their own strategy may reduce gait variability, improve reactions when less common or unique situations arise, and possibly reduce the risk of falls [12], which warrants further studies. On the research side, the results indicate good discriminative validity of our Assessment of Leg Preference in Amputees. It would be of interest to determine the predictive value of inconsistent performance on this instrument in relation to falls.

Limitations

This study has several limitations. Although our sample was larger than in many amputee studies, it still limits generalization of findings. Also, the sample underrepresented amputees due to vascular disease and diabetes. All prosthesis users were rated as K3 or K4, so it is unknown whether our findings translate to the entire population.

References

[1] Peters M. Footedness: asymmetries in foot preference and skill and neuropsychological assessment of foot movement. Psychol Bull. 1988; 103:179-92.

[2] Elias LJ, Bryden MP, Bulman-Fleming MB. Footedness is a better predictor than is handedness of emotional lateralization. Neuropsychologia. 1998; 36:37-43.

[3] Elias LJ, Bryden MP. Footedness is a better predictor of language lateralisation than handedness. Laterality. 1998; 3:41-51.

[4] Pasquina PF, Cooper RA. Care of the Combat Amputee. Textbooks of Military Medicine. Washington, DC: Office of the Surgeon General; 2010. p. 820.

[5] Sadeghi H, Allard P, Prince F, Labelle H. Symmetry and limb dominance in able-bodied gait: a review. Gait Posture. 2000; 12:34-45.

[6] Grouios G. Footedness as a potential factor that contributes to the causation of corn and callus formation in lower extremities of physically active individuals. The Foot. 2005; 15:154-62.

[7] Taylor MJ, Strike SC, Dabnichki P. Turning bias and lateral dominance in a sample of able-bodied and amputee participants. Laterality. 2006; 12:50-63.

[8] Dessery Y, Barbier F, Gillet C, Corbeil P. Does lower limb preference influence gait initation? Gait Posture. 2011; 33:550-5.

[9] Kalaycioglu C, Kara C, Atbasoglu C, Nalcaci E. Aspects of foot preference: differential relationships of skilled and unskilled foot movements with motor asymmetry. Laterality. 2008; 13:124-42.

[10] Schneiders AG, Sullivan J, O'Malley KJ, Clarke SV, Knappstein SA, Taylor LJ. A valid and reliable clinical determination of footedness. PMR. 2010; 2:835-41.

[11] Chen R, Corwell B, Yaseen Z, Hallett M, Cohen LG. Mechanisms of cortical reorganization in lower-limb amputess. J Neurosci. 1998; 18:3443-50.

[12] Miller WC, Speechley M, Deathe AB. The prevalence and risk factors of falling and fear of falling among lower extremity amputees. Arch Phys Med Rehabil. 2001; 82:1031-7.

CHAPTER 4

STRIDE LENGTH-CADENCE RELATIONSHIP IS DISRUPTED IN BELOW-KNEE PROSTHESIS USERS

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10.1016/j.gaitpost.2013.04.008
http://www.sciencedirect.com/science/article/pii/S0966636213001975

Abstract

The aim of this study was to evaluate the linearity of the relationship between stride length and cadence ($STRIDE_{LC}$) over three self-selected speeds (normal, slow, fast) in below-knee prosthesis users (n=14, 11 men, mean age 43±12 years, mean time since amputation 9.2±6.9 years) in comparison to controls (n=20, 11 men, mean age 43±17 years). The step length-cadence relationship ($STEP_{LC}$) was also calculated for the prosthetic and intact legs in prosthesis users and compared to the dominant leg of controls. The goodness of linear fit (R^2) and slope over 3 speeds were used as outcome measures. Prosthesis users walked significantly slower than controls (slow-fast speed means 82-131 vs. 97-169 cm/s, respectively, ANOVA $p<0.0001$) due to both lower cadence (42-53 vs. 47-63 strides/min, $p<0.0001$) and shorter stride length (116-149 vs. 123-161 cm, $p<0.0001$). The R^2 of $STRIDE_{LC}$ relationship in below-knee prosthesis users (0.76±0.13) was significantly lower than in controls (0.91±0.03, $p<0.001$). The R^2 values of $STEP_{LC}$ relationship between the prosthetic and intact legs in prosthesis users were correlated ($r=0.85$, $p<0.001$) and both (0.67±0.19, 0.58±0.21, respectively) were

significantly smaller than in the dominant leg of controls (0.86±0.04, $p<0.01$). The slopes of $STRIDE_{LC}$ and $STEP_{LC}$ were not different. The R^2 of 0.84 for $STRIDE_{LC}$ best discriminated prosthesis users from controls with high sensitivity (71%) and specificity (95%). The results indicate that coupling between stride/step length and cadence is disturbed in prosthesis users. Upon further investigation, the goodness of linear fit may prove to be useful in assessing prosthetic design, optimizing prosthetic fit, and predicting clinical outcomes.

Introduction

Healthy subjects modulate velocity by adjusting both stride length and cadence [1, 2]. Although each of these parameters can be independently modulated, their relationship remains consistent across a wide range of speeds during natural walking [3-5] until it gets disrupted at extreme speeds [2, 3, 6]. This relationship is expressed as a walk ratio (length/cadence) or stride length-cadence plot [5, 6]. The plot of the stride length-cadence relationship follows a close linear pattern across a range of speeds, with similar slopes (walk ratio) in the majority of people without gait impairments [2, 5, 6]. Within-subject consistency of the stride length-cadence relationship over time has also been documented in an unimpaired population [7]. The stride length-cadence relationship, including the walk ratio, has been used to describe pathological gait in Parkinson's patients [6], predict falls in elderly [8, 9], and better understand the neurocontrol of gait in healthy subjects [5, 10].

Characteristics and utility of the stride length-cadence relationship remain unknown in lower-limb prosthesis users. Differences in goodness of fit (R^2) or slope (walk ratio) of the linear relationship between stride length and cadence would indicate changes in the control of gait as a result of amputation or use of a prosthetic device. This has potential clinical and research implications because prosthesis users are at higher risk for falls compared to their peers [11]. Although previous studies suggested that stride length or cadence alone is not useful for predicting falls in prosthesis users [12, 13], their relationship at different speeds was not examined. Considering that stride lengths are shorter at higher cadences in elderly fallers [8, 9], it is plausible that this approach may be more sensitive for predicting falls in prosthesis users. Also, characterizing the stride length-cadence relationship during natural walking would be a step toward validating an assumption that stride length can be derived from cadence when prosthesis users are walking on a treadmill [14].

The objective of this study was to determine if the relationship between stride length and cadence is altered in below-knee prosthesis users. The specific aims were to compare 1) the stride length-cadence ($STRIDE_{LC}$) relationship between below-knee prosthesis users and age-matched controls, and 2) the step length-cadence ($STEP_{LC}$) relationship between the prosthetic limb and intact limb of prosthesis users and the dominant limb of controls. The first hypothesis was that the goodness of linear fit (R^2) of the $STRIDE_{LC}$ relationship would be lower in the below-knee prosthesis users compared with the controls. The second hypothesis was that the R^2 of the $STEP_{LC}$ relationship would be lower in the prosthetic limb than either the intact limb of the prosthesis users or the dominant limb of the controls, with no difference between the intact limb and the control limb. The latter hypothesis was based on the assumption that $STEP_{LC}$ relationship in the intact limb is independent of postulated changes in the prosthetic limb. The slopes (walk ratios) of $STRIDE_{LC}$ and $STEP_{LC}$ relationships were examined in secondary analyses.

Methods

Participants

Unilateral below-knee prosthesis users were recruited from two prosthetic clinics run by our institution. The inclusion criteria were: (1) age 18 – 80 years, (2) comfortable socket fit, (3) no known balance, neurological, or other health problems that limit daily activities, (4) able to safely walk 10m-distance, (5) verification by certified prosthetist that prosthesis user was fit to attempt walking at different velocities. A sample of age-matched able-body control subjects was also recruited with the same relevant criteria.

The study sample included 20 able-bodied controls (11 men; mean age 43±17 years, body mass index 25±3.2) and 14 below-knee prosthesis users (11 men; mean age 43±12 years, body mass index 26±2.6). The average time since amputation was 9.2±6.9 years (range 0.9 to 27.5). The amputation was due to trauma (n=11), infection (n=2), or vascular disease (n=1). Three prosthetic subjects reported having diabetes, but this was not the primary reason for amputation. The prosthetic subjects were rated K3 (n=13) or K4 (n=1) on the Medicare scale. The prosthesis users wore their primary prosthesis and walked without an assistive device. All subjects wore their own shoes during testing. All data were collected by the same researcher at two prosthetic clinics and a hospital's research facility. The study protocol was approved by the institutional review board for human research and all subjects provided informed consent.

Protocol

Temporal and spatial foot fall data were collected while subjects walked over an electronic walkway (GAITRite®, length 5.2 m) at three self-selected speeds (normal, slow, and fast). An additional 1.2 m on each end of the walkway allowed for acceleration and deceleration so that only steady state gait was recorded. Subjects completed a minimum of 6 passes at each speed, which they freely selected in order to achieve the most natural walking pattern. The normal gait speed was always collected first and the order of other two speeds was randomized across subjects. Demographic and clinical information were collected through an interview and chart review.

Data processing

The collected foot fall data were processed with a custom program written in MATLAB® (Mathworks Inc., Natick, MA) to derive stride velocity (cm/s), stride length (cm), instantaneous stride cadence (strides/min), step length (cm), and instantaneous step cadence (steps/min). Instantaneous stride and step cadence were calculated from the individual stride and step times. Stride parameters were calculated when the dominant or prosthetic side was the lead foot.

For evaluation of the $STRIDE_{LC}$ relationship, each foot fall was treated as an individual data point. The linear regression was derived from all stride length-stride cadence pairs across the three speeds to compare the prosthesis users to controls (hypothesis 1). Identical analyses were conducted for step length-step cadence pairs for comparison between the prosthetic limb, intact limb, and the dominant limb of the controls (hypothesis 2). Figure 1 illustrates examples of the $STRIDE_{LC}$ relationship for a representative control subject and a prosthesis user. The coefficient of determination (R^2)

of the regression line was used to evaluate the goodness of fit and served as the main outcome measure for testing the two hypotheses. The slopes of the regression lines were secondary outcome measures.

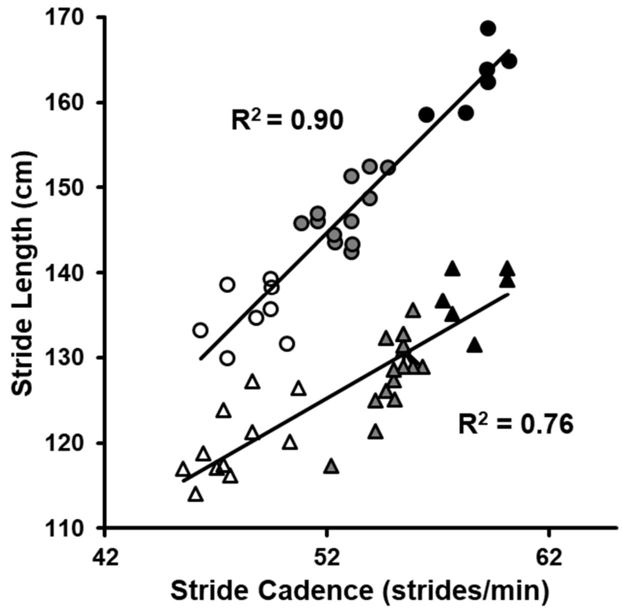

Fig. 1. Examples of the stride length-cadence linear regression for a representative control subject (circles) and prosthesis user (triangles) across slow (white), normal (gray), and fast (black) self-selected speeds. The R^2 value indicates the goodness of fit.

Statistical analysis

For descriptive purposes, the stride and step parameters for each speed were compared between the two groups. A two-way ANOVA (α=0.05) with leg and speed as the main factors was used to evaluate each stride and step parameter. Paired and unpaired t-tests were used on stride and step parameters as appropriate to test differences between the prosthetic, intact, and the dominant leg in controls across different speeds. For these analysis, the significance level was reduced to adjust for multiple comparisons (α=0.025 and α=0.0167 respectively).

To test the first hypothesis, the R^2 values from the $STRIDE_{LC}$ relationship were compared between the control subjects and prosthesis users by an unpaired t-test with a significance level of 0.05. For the second hypothesis, the R^2 values from the $STEP_{LC}$ relationship were submitted to a one-way ANOVA (α=0.05) to test differences between the prosthetic, intact, and dominant leg in controls. If the main effect of leg was significant, Tukey's HSD test was conducted between each pair. Pearson's correlation (r) was used to assess the relationship between the R^2 values in the prosthetic and intact legs. The slopes from the $STRIDE_{LC}$ and $STEP_{LC}$ relationships were similarly compared with the unpaired t-test and one-way ANOVA, respectively. Gait parameters in the prosthesis users were correlated with age and time since amputation to examine potential confounds.

Since the R^2 value of $STRIDE_{LC}$ relationship was found to be significantly different between the control and prosthesis user groups, post-hoc analysis was conducted to determine the sensitivity and specificity of this measure. A receiver operating curve was used to identify the cutoff point that best discriminates prosthesis users from controls. The R^2 value with the best likelihood ratio was chosen as the cutoff point. Prism 5 software (GraphPad, La Jolla, CA) was used for statistical analysis.

Results

Comparison of gait parameters across 3 speeds

Two-way ANOVA revealed significant main effects of leg and speed for all gait parameters without significant interactions (Table 1). The prosthesis users walked consistently slower than controls due to significant reduction in both stride length and cadence. Both prosthesis users and controls made comparable adjustments when asked to walk slower (mean -24% vs. -28%) and faster (+21% vs. +26%) than normal self-selected speed. The main effect of speed affirms that the subjects complied with the request to modulate the walking speed. The lack of significant leg x speed interaction indicates that the gait parameters were modulated at a comparable rate between the two groups.

Further t-test comparisons revealed that the stride velocity was slower and the stride cadence was lower in the prosthesis users than controls at the normal and fast speeds (Table 1). Step length was significantly shorter in the intact leg of prosthesis users compared to the dominant leg of controls at the normal and fast speeds. Also, step cadence was bilaterally lower in the prosthesis users compared to controls for all speeds, except for the intact leg at the slow speed. In comparison to the intact leg, the step length in the prosthetic leg was significantly longer at normal and fast speeds, whereas the step cadence was significantly lower at the slow speed only.

Stride length-cadence relationship

The R^2 value of the $STRIDE_{LC}$ relationship was significantly lower in the prosthesis users (0.76±0.13) than the control subjects (0.91±0.03, $p<0.001$), which confirmed the first hypothesis. Such large differences between the prosthesis users and controls are evident in Figure 2, which shows the individual R^2 values and the group means. In contrast to the R^2 values, the slopes of the $STRIDE_{LC}$ relationship were not significantly different between prosthesis users and controls (2.9±1.2 vs. 2.4±0.6, respectively, $p=0.15$).

Table 1
Mean (SD) values for stride and step parameters at the three speeds with the ANOVA results.

	Slow	Normal	Fast	ANOVA p-values		
				Leg	Speed	Leg x Speed
Stride Velocity (cm/s)				<0.0001	<0.0001	0.160
Control	97 (20)	134 (17)	169 (24)			
Prosthetic	82 (16)	108 (14)*	131 (21)*			
Stride Length (cm)				0.004	<0.0001	0.623
Control	123 (14)	144 (12)	161 (12)			
Prosthetic	116 (15)	133 (15)	149 (19)			
Stride Cadence (strides/min)				<0.0001	<0.0001	0.335
Control	47 (6)	55 (4)	63 (7)			
Prosthetic	42 (5)	49 (4)*	53 (5)*			
Step Length (cm)				0.006	<0.0001	0.936
Control	62 (7)	72 (7)	80 (6)			
Prosthetic	59 (10)	69 (9)+	77 (12)+			
Intact	56 (7)	64 (7)*	72 (9)*			
Step Cadence (steps/min)				<0.0001	<0.0001	0.485
Control	94 (11)	111 (9)	125 (14)			
Prosthetic	84 (10)*	96 (8)*	105 (11)*			
Intact	86 (8)+	99 (7)*	108 (9)*			

* significant un-paired t-test, p≤0.0167
+ significant paired t-test, p≤0.025

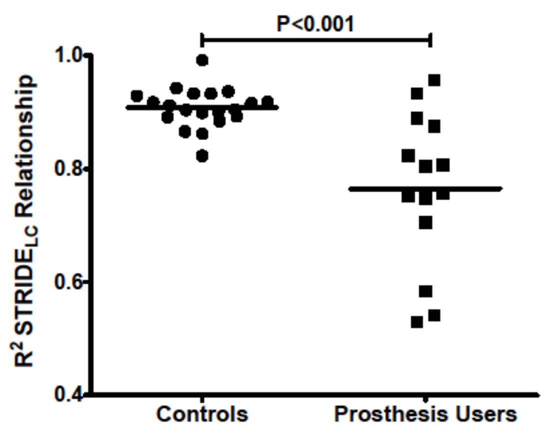

Fig. 2. Individual and mean R^2 values of the stride length-cadence relationship for control subjects (circles) and prosthesis users (squares). Note that only 4 prosthesis users overlap with controls.

The R^2 value of 0.84 best discriminated prosthesis users from controls (likelihood ratio 14.29), with a sensitivity of 71% and specificity of 95% (Figure 3). The area under the receiver operating curve was 0.85 (confidence interval 0.69-1.00, $p<0.001$). With the cutoff of 0.84, 19 of the 20 (95%) controls were considered within normal limits but only 4 of the 14 (29%) prosthesis users.

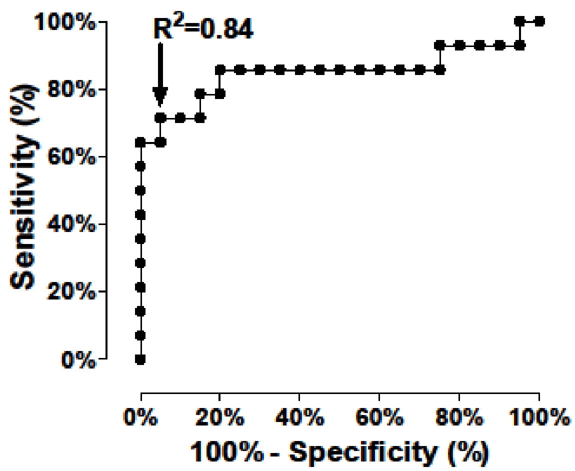

Fig. 3. The receiver operating curve based on R^2 values of the stride length-cadence relationship. An R^2 of 0.84 best discriminated prosthesis users from controls (71% sensitivity, 95% specificity).

Step length-cadence relationship

The R^2 values of the $STEP_{LC}$ relationship were significantly different between the prosthetic, intact, and control legs (ANOVA $p<0.0001$). Tukey's comparison revealed that the dominant leg of controls (0.86±0.04) had significantly higher R^2 values than both the prosthetic (0.67±0.19, $p<0.01$) and intact legs (0.58±0.21, $p<0.001$), with no difference between the latter two. The R^2 values for the prosthetic and intact legs of prosthesis users strongly correlated with each other ($r=0.85$, $p<0.001$). These results only partially confirmed the second hypothesis because, instead of the postulated difference between the prosthetic and intact legs, we found that both legs of the prosthesis users differed from controls. This is illustrated in Figure 4, which shows the individual and group mean data for the controls and each leg of the prosthesis users.

As with the stride data, the prosthetic, intact, and dominant legs did not differ in terms of the slope (0.60±1.15, 0.67±0.27, 0.63±0.37, p=0.78) of the $STEP_{LC}$ relationship. Age and time since amputation did not significantly correlate with any parameter of the $STRIDE_{LC}$ or $STEP_{LC}$ relationship.

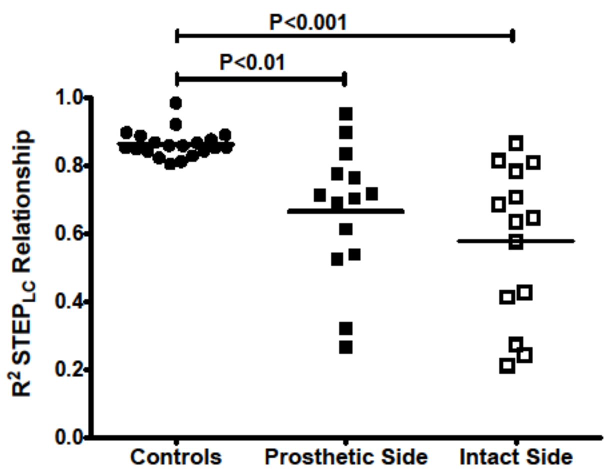

Fig. 4. Individual and mean R^2 values of the step length-cadence relationship for the dominant leg of controls (circles) and the prosthetic (filled squares) and intact legs (open squares) of prosthesis users. Note that both legs of prosthesis users are significantly different from controls but not between each other.

Discussion

This study evaluated the linear relationship between the stride length and stride cadence and the step length and step cadence in below-knee prosthesis users in comparison with age-matched controls. The results indicate that the goodness of linear fit (R^2) between stride/step length and cadence is lower in both legs of below-knee prosthesis users compared with the dominant leg of controls. No significance for the slope (walk ratio) suggests that prosthesis users modulate stride/step length and cadence no differently than controls when changing speeds. The R^2 value of 0.84 from $STRIDE_{LC}$ relationship effectively discriminates prosthesis users from controls. The R^2 for the $STEP_{LC}$ relationship in prosthesis users significantly correlates between the prosthetic and intact legs. The findings are unrelated to age or time since amputation.

The sample recruited for this study seems representative of ambulatory below-knee prosthesis users. Their demographic and basic gait characteristics are similar to prosthesis users evaluated in other studies [13, 15]. Despite differences in gait between our prosthesis users and controls, both groups comparably modulated velocity when asked to walk slower and faster than the normal speed (about ±25%). This is confirmed by ANOVA as the lack of leg x speed interaction for various gait parameters. No significant interaction indirectly reflects the lack of difference between the two groups in the slope of stride\step length-cadence relationship.

To the best of our knowledge, no previously published study examined coupling between the stride/step length and cadence in below-knee prosthesis users. Although not explicitly studied, weaker coupling between the stride length and cadence was apparent in below-knee prosthesis users walking in tall grass [16]; whereas controls proportionally reduced both stride length and cadence, prosthesis users only reduced stride length. However, a disruption of the linear relationship has been reported in able-bodied adults when stride length or cadence is restricted [5]. By analogy, our results may imply that the prosthetic device imposes a constraint on the sensori-motor control of gait reflected by reduced linearity between stride/step length and cadence.

The key finding of this study is a weaker coupling between the stride/step length and cadence across comfortable self-selected speeds in prosthesis users. No difference in the slope suggests that prosthesis users are capable of modulating the coupling between stride\step length and cadence across various speeds. The fact that the $STEP_{LC}$ relationship is bilaterally disrupted and exhibited high correlation between the prosthetic and intact sides provides evidence that the disruption in the $STRIDE_{LC}$ relationship is due to a bilateral loss of coupling. The apparent correlation in coupling between the two legs is in contrast to an asymmetric gait pattern seen in our prosthesis users who walked with shorter steps and faster cadence on the intact side. The follow-up analyses argue against a possibility that the disruption of the stride/step length-cadence relationship in prosthesis users was due to a greater scatter of data points at the lower speed. Also, no other model fit the data better than the linear regression line. This confirms that the decoupling of the stride/step length-cadence relationship is an inherent property of below-knee prosthetic gait. Thus, the main findings likely reflect different aspects of altered neurocontrol of gait in prosthesis users.

The potential causes of the disrupted stride/step length-cadence relationship in below-knee prosthesis users are not apparent at this time. The basic gait parameters and the type of prosthetic device in the four subjects with R^2 values above the 0.84 cutoff point were not substantially different from the rest of the prosthetic population. Age and time since amputation also did not play a major role, although our sample size may be too small for accurate assessment. Other potentially contributing factors that need to be examined in the future include the type and duration of gait training, alignment, foot type, energy return properties of the prosthetic device, and the level of comfort within the socket.

The implications of the weaker linear relationship between stride/step length and cadence in prosthesis users also needs further studies. We postulate that weaker coupling between these parameters across the range of speeds may be related to a fall risk. This method for assessing fall risk is different from conventional approaches that are based on variability of selected gait parameters at distinct speeds. Previous studies in elderly populations reported differences in the walk ratio between fallers and non-fallers only at higher speeds, but the strength of linear relationship was not examined [8, 9]. Since the walk ratio (slope) was not significantly different between the two groups in our study, we suspect that it is unlikely to be a good predictor of fall risk in prosthesis users.

Conclusion

Evaluation of the $STRIDE_{LC}$ relationship has an ecological validity because it captures key characteristics of gait in the manner that resembles everyday life since prosthesis users are expected to walk at different speeds depending on the environment or situation. Since both stride length and cadence are strongly linked to velocity, even slight changes in velocity require concomitant and proportional adjustments in both stride length and cadence. Thus, disturbed coupling between stride\step length and cadence in prosthesis users may better represent gait deviations as they occur in natural settings than when studied at individual speeds. Better understanding of $STRIDE_{LC}$ relationship also has implications for research studies utilizing a treadmill. Based on the assumption that this relationship is preserved in prosthesis users, it is customary to derive stride length from cadence and velocity of the treadmill. However, this practice is questioned based on the evidence of disrupted relationship between the stride/step length and cadence in below-knee prosthesis users that was observed in this study. Finally, our first estimate of the $STRIDE_{LC}$ cutoff point (R^2=0.84) that adequately discriminates below-knee prosthesis users from controls may serve to track the progress of rehabilitation and assess how well attained results approximate walking of unimpaired subjects.

Limitations

This study has several limitations. Although larger than in many other studies of amputee gait, the sample size of below-knee prosthesis users is relatively small. None of the prosthesis users used an assistive device, which limits generalization to those who walk with an assistive device. Also, a variety of prosthetic devices were used and it remains unknown whether that confounded the results.

References

[1] Murray MP, Kory RC, Clarkson BH, Sepic SB. Comparison of free and fast speed walking patterns of normal men. PMR. 1966; 45:8-24.

[2] Grieve DW, Gear RJ. The relationships between length of stride, step frequency, time of swing and speed of walking for children and adults. Ergonomics. 1966; 5:379-99.

[3] Du Chatinier K, Molen NH, Rozendal RH. Step length, step frequency and temporal factors of the stride in normal human walking. Anatomy. 1970; 73:214-27.

[4] Andriacchi TP, Ogle JA, Galante JO. Walking speed as a basis for normal and abnormal gait measurments. J Biomech. 1977; 10:261-8.

[5] Egerton T, Danoudis M, Huxham F, Iansek R. Central gait control mechanisms and the stride length - cadence relationship. Gait Posture. 2011; 34:178-82.

[6] Morris M, Iansek R, Matyas T, Summers J. Abnormalities in stride length-cadence relation in parkinsonian gait. Mov Disord. 1998; 13:61-9.

[7] Sekiya N, Nagasaki H. Reproducibility of the walking patterns of normal young adults: test-retest reliability of the walk ratio(step-length/step-rate). Gait Posture. 1998; 7:225-7.

[8] Barak Y, Wagenaar RC, Holt KG. Gait characteristics of elderly people with a history of falls: a dynamic approach. Physical Therapy. 2006; 86:1501-10.

[9] Callisaya ML, Blizzard L, McGinley JL, Srikanth VK. Risk of falls in older people during fast-walking - The TASCOG study. Gait Posture. 2012; 36:510-5.

[10] Zijlstra W, Rutgers AWF, Hof AL, Van Weerden TW. Voluntary and involuntary adaptation of walking to temporal and spatial constraints. Gait Posture. 1995; 3:13-8.

[11] Miller WC, Speechley M, Deathe AB. The prevalence and risk factors of falling and fear of falling among lower extremity amputees. Arch Phys Med Rehabil. 2001; 82:1031-7.

[12] Parker K, Hanada E, Adderson J. Gait variability and regularity of people with transtibial amputations. Gait Posture. 2013; 37:269-73.

[13] Vanicek N, Strike S, McNaughton L, Polman R. Gait patterns in transtibial amputee fallers vs. non-fallers: Biomechanical differences during level walking. Gait Posture. 2009; 29:415-20.

[14] Houdijk H, Pollmann E, Groenewold M, Wiggerts H, Polomski W. The energy cost for the step-to-step transition in amputee walking. Gait Posture. 2009; 30:35-40.

[15] Isakov E, Burger H, Krajnik J, Gregoric M, Marincek C. Influence of speed on gait parameters and on symmetry in transtibial amputees. Prosthet Orthot Int. 1996; 20:153-8.

[16] Paysant J, Beyaert C, Datie AM, Martinet N, Andre JM. Influence of terrain on metabolic and temporal gait characteristics of unilateral transtibial amputees. J Rehabil Res Dev. 2006; 43:153-60.

CHAPTER 5

INCREASED ALERTNESS, BETTER THAN POSTURE PRIORITIZATION, EXPLAINS DUAL-TASK PERFORMANCE IN PROSTHESIS USERS AND CONTROLS UNDER INCREASING POSTURAL AND COGNITIVE CHALLENGE

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Howard CL, Perry B, Chow JW, Wallace C, Stokic DS. Increased alertness, better than posture prioritization, explains dual-task performance in prosthesis users and controls under increasing postural and cognitive challenge. Exp Brain Res. 2017; Epub: Aug 31; 10.1007/s00221-017-5077-2.
https://link.springer.com/article/10.1007/s00221-017-5077-2

Abstract

Sensorimotor impairments after limb amputation impose a threat to stability. Commonly described strategies for maintaining stability are the posture first strategy (prioritization of balance) and posture second strategy (prioritization of concurrent tasks). The existence of these strategies was examined in 13 below-knee prosthesis users and 15 controls during dual-task standing under increasing postural and cognitive challenge by evaluating path length, 95% sway area, and anterior-posterior and medial-lateral amplitudes of the center of pressure. The subjects stood on two force platforms under usual (hard surface/eyes open) and difficult (soft surface/eyes closed) conditions, first alone and while performing a cognitive task without and then with instruction on cognitive prioritization. During standing alone, sway was not significantly different between groups. After adding the cognitive task without prioritization instruction, prosthesis users increased sway more under the dual-task than single-task standing

($p \leq 0.028$) during both usual and difficult conditions, favoring the posture second strategy. Controls, however, reduced dual-task sway under a greater postural challenge ($p \leq 0.017$), suggesting the posture first strategy. With prioritization of the cognitive task, sway was unchanged or reduced in prosthesis users, suggesting departure from the posture second strategy, whereas controls maintained the posture first strategy. Individual analysis of dual-tasking revealed that greater postural demand in controls and greater cognitive challenge in prosthesis users led to both reduced sway and improved cognitive performance, suggesting cognitive-motor facilitation. Thus, activation of additional resources through increased alertness, rather than posture prioritization, may explain dual-task performance in both prosthesis users and controls under increasing postural and cognitive challenge.

Introduction

Postural control is maintained actively and passively through coordinated responses to visual, vestibular, and somatosensory inputs, the mechanical support provided by the musculoskeletal system, and involvement of cognitive resources [1, 2]. The contributions of these control systems may be reduced due to environmental demands, physical and cognitive limitations, or multi-tasking, which requires the remaining control systems to take on a greater role in maintaining balance [2, 3]. With more strain on these control systems, balance performance may degrade [1]. Many studies illustrate detriments in balance when sensory feedback loops are altered or impaired [4]. In contrast, individuals without impairments maintain body sway even when faced with an additional postural challenge [1, 5, 6]. This capacity to adjust to increased postural demand is referred to as postural reserve [3].

The sensory, mechanical, and cognitive systems contributing to the maintenance of posture are also involved in many other tasks. With ample resources available, multi-tasking does not negatively affect balance or performance on competing tasks. However, with increasing task difficulty and available resources depleted, successful performance on one task may require shifting resources away from the other tasks, depending on their priority [1, 3, 7]. When performance on a task is sacrificed in favor of maintaining balance, this strategy is referred to as the "posture first strategy" [8]. Conversely, the "posture second strategy" is when balance is sacrificed in favor of the other tasks, which may pose a risk to stability [2, 3, 5].

Use of the posture first or posture second strategy has been assessed with a dual-task paradigm, which involves performing an additional cognitive or motor task while standing or walking. As the dual-task requirements become more challenging, individuals without impairments tend to follow the posture first strategy, whereas persons with sensorimotor impairments often do not [3, 5, 8-11]. The latter observations mainly come from persons with Parkinson's disease or stroke [7, 9, 12], making it difficult to disentangle possible contributions of disequilibrium, weakness, altered muscle tone, or cognitive deficits on the choice of postural strategy.

Lower-limb amputees suffer from partial losses to musculoskeletal, motor, and somatosensory systems, which affect postural control. Despite improvements in prosthetic designs, prosthesis users remain at increased risk for falls [13, 14]. Minor threats to stability are presumably compensated for by the remaining resources in the postural reserve available to prosthesis users [15]. The use of the postural reserve could explain near normal sway characteristics reported in prosthesis users during normal standing [16, 17]. However, increasing challenge may deplete limited postural resources and result in situations where not all demands can be fully met. Prosthesis users may follow the posture first strategy to accommodate for the reduced postural reserve, which is supported by the findings of little to no increase in unsteadiness when prosthesis users are asked to concurrently perform a cognitive task, despite self-reports of increased cognitive burden [18, 19]. On the other hand, there are reports of increased unsteadiness with dual-tasking [20, 21], suggestive of the posture second strategy. Some of the existing controversy may be due to the differences in methodology and studied population (above- vs. below-knee prosthesis users), and not accounting for performance on a concurrent task. The use of the posture second strategy in prosthesis users could provide an explanation for greater fall risk and point to approaches for reducing the risk for falls in this population.

To determine if the posture first or posture second strategy is used during dual-tasking, it is necessary to impose experimental conditions that stress the postural reserve. The stress should be sufficient enough to tap into the postural reserve and force a reallocation of resources between the competing tasks. In order to determine how amputation impacts strategy of choice in balance maintenance, we combined a dual-task paradigm with challenging postural tasks in the evaluation of prosthesis users and age- and education-matched non-amputee controls. Our rationale was that both prosthesis users and controls may initially allow some sacrifices in postural control in favor of better performance on a cognitive task (posture second strategy) when the risk to stability is low. However, when the postural demand rises, more resources may be allocated to posture to limit unsteadiness, which comes at the expense of cognitive performance (posture first strategy). This will manifest as a smaller increase in sway in dual-task than single-task standing under greater postural challenge. In line with the posture first strategy, this shift in resources to maintain postural control would also result in worse cognitive performance [3]. Thus, to examine how a change in sway is related to allocation of available resources, the performance on the concurrent cognitive task should also be considered [22]. This is expected to provide a better appreciation of the cognitive-motor interaction, which may yield not only cognitive-motor interference but also facilitation.

Based on this rationale, this study was designed with two major aims. Our first aim was to examine changes in sway under increasing postural challenge between single-task and dual-task conditions without specific instructions on prioritization. The following hypotheses were tested regarding the first aim. Since more challenging standing conditions are needed to perturb balance in prosthesis users [16, 17], we postulated that the sway in prosthesis users will be no different from controls under the usual single-task standing condition (hypothesis 1). As a corollary to this hypothesis, we predicted that with greater postural challenge in single-task standing both groups will increase sway, but the increase will be greater in prosthesis users than controls (hypothesis 1A). In terms of the strategy used by each group, we hypothesized that with greater postural challenge both groups will follow the posture first strategy, such that the increase in sway during dual-task standing will be *smaller* than the increase in sway during single-task standing (hypothesis 2). Failure to follow this behavior suggests the posture second strategy.

In our second aim, we examined the impact of instruction to focus on improving cognitive performance during dual-task standing (cognitive task prioritization) to assess if overt cognitive prioritization can impact the postural strategy in each group [9, 23]. We hypothesized that the instruction to improve cognitive performance will result in increased sway dual-task cost from the no-prioritization to prioritization condition, suggesting the posture second strategy (hypothesis 3). However, if the cognitive prioritization instruction does not further disrupt sway (i.e., insignificant change or significant reduction), the posture first strategy or mutual facilitation is implicated. Thus, the relationships between sway dual-task cost and cognitive dual-task cost were also assessed under different conditions to infer changes in strategy at the group and individual levels in terms of all possible outcomes of cognitive-motor interaction (i.e., interference vs. facilitation) [22].

Methods

Participants

A convenience sample of unilateral below-knee prosthesis users (n=16) was recruited from our clinics by a certified prosthetist. The inclusion criteria were ≥ 1 year since amputation; age 18–80 years; comfortable socket fit; no known balance, neurological, or other health problems that limit daily activities; and able to safely stand without the use of an assistive device. Age- and education-matched non-amputee controls were recruited from the community (n=17). The study was approved by the institutional review board for human research and all subjects signed the informed consent.

Protocol

General cognitive functions were assessed using the Modified Mini-Mental Status Exam (3MS) and processing speed and executive function by Trail-Making (Trail) forms A and B (scaled T-scores adjusted for race, age, gender, and education; higher scores represent better performance). Demographic and clinical data were collected through an interview and from medical records.

All standing tasks were performed using 2 force plates (Type 4060, 40x60 cm^2, Bertec Corp, Columbus, OH). Subjects were instructed to place each foot on a separate force plate with shoes on, stand naturally with a shoulder width stance, and keep arms comfortably and freely at their sides. Foot placement was marked to ensure consistent standing position across all trials. Force plate data were collected using a Cortex data acquisition system (Motion Analysis Corp., Santa Rosa, CA, sample rate 1,200 Hz, 12-bit analog-to-digital resolution).

Two standing surfaces (hard and soft) and 2 vision conditions (eyes open and closed) were used to create increasing levels of postural challenge. This produced 4 standing conditions with hard surface/eyes open considered the least challenging (referred to as usual standing) and the soft surface/eyes closed considered the most challenging (difficult standing). The intermediate conditions (provided in Appendix A) were not reported here to emphasize extreme effects. The soft standing condition required subjects to stand with each foot on a foam pad (Airex Balance Pad, Sins, Switzerland, 40x50x6 cm^3, 0.726 kg, density 61 kg/m^3). The pads were positioned so that they did not touch each other or extend over the edge of the force plate.

The choice of instructions and order of their presentation were selected after extensive pilot testing. All single-task standing conditions were collected prior to the dual-task conditions. Subjects performed two 30-s trials for each surface/vision combination. The hard surface was collected first to familiarize subjects with the procedure. Eyes open was collected first in each surface condition. Subjects were allowed to take seated breaks as needed and step off the force plates or foam pads between tasks.

Two cognitive tasks were selected for the dual-task paradigm; serial subtraction by 7 from a 3-digit number and a verbal fluency task (listing words starting with a specific letter). The most difficult letters for verbal fluency (J, K, Q, U, X, Y, Z) were excluded from this task [24]. Each task was practiced while seated to ensure comprehension. The subtraction task was performed 2-3 times for 30 s as a seated baseline performance. The verbal fluency F-A-S test (FAS) was performed once for 60 s of which the standard 60-s score was used for comparisons to controls and the first 30-s score as a seated baseline performance. The number of correct responses was documented, and the verbal responses were also recorded to confirm response accuracy.

After completing the single-task standing condition and a brief seated rest, subjects repeated each surface/vision combination while performing each cognitive task once for 30 s (dual-task standing). The order of the 2 cognitive tasks and 2 surface conditions was randomized with eyes open always collected first for each surface. In aim 1, subjects were given no instruction on task prioritization. In aim 2, during the dual-task standing, subjects were asked to focus on the cognitive task and increase the number of correct responses by at least 50% over their noted average in aim 1. In each aim, an additional subtraction task was given at random as a distractor (subtracting 6 or 8, data not included). The performance on each cognitive task was documented and recorded.

Data processing

Ground reaction forces and moments were used to determine the center of pressure (CoP) of each foot contact. The resultant CoP is a point along the line connecting the 2 CoPs and the location was determined using the equation of equilibrium (i.e., the sum of the moments due to individual vertical ground reaction forces about the resultant CoP equaled to zero). The resultant CoP locations were then filtered with a 4th order low-pass Butterworth filter (cutoff frequency 10 Hz).

The primary outcome measures were path length (PL, the sum of the distance between adjacent resultant CoP locations) and sway area (AREA, the best fit ellipse that captures 95% of the resultant CoP locations). After computing PL from both the original data sampled at 1200 Hz and the data down-sampled to 200 Hz, we found consistently longer PL values for the 1200-Hz than the 200-Hz data. To ensure the reported results are comparable to the literature, the down-sampled data were used. The ranges of resultant CoP locations in the anterior-posterior and medial-lateral directions (AP and ML amplitudes, respectively) were used as secondary outcome measures to determine if changes in sway were driven by movement in a specific direction. All computations were completed using a custom program written in MATLAB® (Mathworks Inc., Natick, MA). Due to sporadic artifacts at the beginning or end of some trials, the middle 5,120 samples (25.6-s of the 30-s trial) were analyzed for consistency between subjects. The data of 3 prosthesis users and 1 control subject were excluded due to technical problems or a violation of the protocol. After data inspection, 1 control subject was considered an outlier (50% of outcomes exceeded 3 SD of the group mean) and excluded from further analysis.

Due to personal aptitude, the same task may not have equal interfering effect in all subjects. Since our goal was to examine the change in postural strategy under an undoubtedly stressful condition, the cognitive task (subtraction or verbal fluency) with a more disruptive effect across the 4 sway parameters was identified in each subject and selected for the dual-task analysis (only 1 of the 2 cognitive tasks was available for analysis in 1 prosthesis user due to technical issues). The same approach was used in our previous study [20]. The distribution of the subtraction and verbal fluency tasks was not significantly different between the two groups (Fisher's exact test p=0.5). Dual-task cost was calculated as the difference between the single-task and dual-task standing for the 4 sway parameters and the cognitive performance on a more disruptive task across all surface/vision combinations and prioritization conditions (negative sign indicates greater sway/worse cognitive performance during dual-task standing).

To appreciate the strategy employed, changes in both sway and cognitive performance should be considered [22]. Thus, for both prioritization conditions, the dual-task cost for PL and AREA was plotted against the respective cognitive dual-task cost to infer if one, both, or neither task was affected by the concurrent performance or if they were affected in different ways. For that, the plot area was divided into regions of interference (negative dual-task cost), no influence, and facilitation (positive dual-task cost) along each axis, resulting in a 3x3 matrix of cognitive-motor interaction. The no influence region for the sway parameters was bounded by the average standard deviation of all single-task standing conditions across both groups (PL ±5.5 cm, AREA ±1.3 cm^2).

For the cognitive task, the no influence region was delimited by ±1 point, after rounding the average standard deviation of the correct responses across both groups (1.1±0.9). Given the emphasis in hypothesis testing on the presence of interference effect, the no influence regions were combined with the respective facilitation regions to reduce the 3x3 matrix to a 2x2 matrix with the following 4 regions: posture facilitation/cognitive interference (posture first strategy), posture interference/cognitive facilitation (posture second strategy), mutual interference, and no change/mutual facilitation [9, 22]. Within each group, the number of subjects in each region was tallied for the usual and difficult standing conditions.

Statistical analysis

Baseline cognitive performance was compared between prosthesis users and controls (unpaired t-test, $p \leq 0.05$). Hypothesis 1, of no difference in sway parameters between the two groups under the usual single-task standing condition, was tested using the average of the two baseline conditions (unpaired t-tests, $p \leq 0.05$). To reveal the greatest impact of a postural challenge, the reported analysis of sway data was limited to the two extreme standing conditions (usual, difficult). Therefore, to test whether increasing the postural challenge affected sway parameters differently in the two groups (hypothesis 1A), a 2x2 mixed ANOVA was used with Group (prosthesis users, controls) as the between-subjects factor and Standing condition (usual, difficult) as the within-subjects factor. The effect of increasing the postural challenge was determined by the main effect of Standing and the differential response of the two groups by the Group x Standing interaction or the main effect of Group ($p \leq 0.05$).

In order to assess the strategy employed by each group during the dual-task standing, we used a within group 2x2 repeated measure ANOVA with Task (single, dual) and Standing (usual, difficult) as factors. Hypothesis 2 was accepted if there was a significant Task x Standing interaction consistent with a *smaller* increase in dual-task sway compared to single-task sway with increasing postural challenge, suggesting the posture first strategy. The posture second strategy was implicated by a significant main effect of Task or Task x Standing interaction due to a *greater* increase in dual-task vs. single-task sway ($p \leq 0.05$).

To assess in each group if the instruction to prioritize the cognitive task led to the posture second strategy, sway dual-task cost was submitted to a 2x2 repeated measure ANOVA with Instruction (no-prioritization, prioritization) and Standing (usual, difficult) as factors. Hypothesis 3 was accepted in case of a significant main effect of Instruction or significant Instruction x Standing interaction, supporting an increase in sway under the prioritization condition ($p \leq 0.05$).

The cognitive dual-task cost was evaluated using a 2x2x2 mixed ANOVA with Group (prosthesis users, controls) as the between-subjects factor and Standing condition (usual, difficult) and Instruction (no-prioritization, prioritization) as the within-subjects factors ($p \leq 0.05$).

Finally, to determine how individual subjects in each group changed the strategy under increasing postural and cognitive challenge, we submitted categorical frequency distributions of the 4 regions of the plot defined by the sway dual-task cost (PL, AREA) vs. cognitive dual-task cost to a general linear model with repeated measures on Standing (usual, difficult) and Instruction (no-prioritization, prioritization) factors ($p \leq 0.05$). The statistics of interest was the significance of either of the two main effects or of their interaction. The α level was set at 0.05 for all tests. No adjustment for cognitive function was made because the two groups did not differ on standard tests of cognitive function (3MS, Trails A/B, FAS). IBM SPSS Statistics 23 (IBM Corp., Armonk, NY) was used for statistical analysis.

Results

The studied sample included 15 controls (mean age 49±16 years, 15±2 years of education, BMI 29.7±7 kg/m^2, 7 (47%) men) and 13 below-knee prosthesis users (age 46±11 years, 14±3 years of education, BMI 31.4±6 kg/m^2, 9 (70%) men). The amputation occurred 8±7 years earlier (range 1.0 to 22 years) due to trauma (n=6), infection (n=2), or vascular disease (n=5). Individual subject characteristics are provided in Table 1. All below-knee prosthesis users were rated K3 on the Medicare scale and none used an assistive device. They all used an energy storage and return style foot, which also included a hydraulic ankle in 3. Two used a passive suction suspension system, 7 elevated vacuum, and 4 a pin locking system. Nine prosthesis users reported living an active or very active lifestyle, 3 reported moderate activity, and 1 sedentary.

The baseline cognitive performance did not significantly differ between prosthesis users and controls (3MS 95±5 vs. 95±4; Trail A 48±10 vs. 44±13; Trail B 48±13 vs. 48±12, FAS 35.6±8 words vs. 39.1±13 words respectively, $p>0.4$ for all measures; for individual scores, see Table 1). The 5 vascular disease amputees did not differ from the controls or other prosthesis users ($p>0.4$).

Table 1

Demographic characteristics and baseline cognitive performance of enrolled subjects.

Subject	Age (years)	Education (years)	Gender	BMI (kg/m^2)	Time since Amputation (years)	Etiology	3MS	Trail A	Trail B	FAS (words)	Cognitive Task
Prosthesis Users											
P01	34	12	M	25.9	14.8	Trauma	100	44	48	42	VF
02	34	17	F	26.6	16.0	Vascular	99	40	55	48	VF
P03	36	14	M	45.0	1.0	Vascular	97	59	68	30	SUB
P04	37	12	M	35.7	4.3	Infection	97	56	68	36	SUB
P05	37	14	F	25.8	4.0	Trauma	95	46	47	20	SUB
P06*	40	17	F	33.5	2.6	Trauma	88	41	40	23	NA
P07	43	7	M	32.4	2.6	Trauma	85	35	17	31	VF
P08	43	16	F	38.0	22.0	Trauma	99	28	41	24	SUB
P09	45	12	M	31.7	4.3	Trauma	100	57	49	41	VF
P10	45	14	M	36.6	6.0	Infection	98	51	51	32	SUB
P11^	50	16	F	23.0	3.5	Infection	100	26	40	39	NA
P12	50	16	F	21.9	18.0	Trauma	93	62	44	49	VF
P13*	55	8	M	32.0	3.0	Vascular	56	27	29	1	NA
P14	59	14	M	32.3	1.8	Vascular	95	49	39	36	VF
P15	62	14	M	25.7	1.0	Vascular	89	39	40	38	VF
P16	71	14	M	30.1	5.0	Vascular	90	57	52	36	SUB
Controls											
C01	30	18	M	25.8			99	35	41	58	SUB
C02	31	18	M	22.7			100	48	45	39	SUB
C03	32	16	F	21.8			89	36	55	41	VF
C04	33	16	F	23.6			97	39	46	43	SUB
C05	37	16	F	28.3			99	60	56	41	SUB
C06	40	14	F	36.9			99	18	39	26	SUB
C07	40	15	M	29.8			92	52	49	19	VF
C08	45	14	F	46.6			90	36	32	28	VF
C09	57	13	F	34.2			96	48	55	43	SUB
C10	57	16	F	38.7			97	50	51	36	SUB
C11	58	13	M	32.1			95	29	47	23	SUB
C12^	60	10	F	30.2			91	29	39	30	NA
C13	64	17	M	22.5			100	66	68	72	SUB
C14	67	12	M	25.1			99	57	70	44	VF
C15	67	14	M	27.0			88	45	26	43	VF
C16†	73	16	M	26.5			99	43	46	32	NA
C17	81	14	F	30.0			93	38	39	30	VF

Subjects not included in analysis: ^ technical issue; * violation of protocol; † outlier.
3MS, Modified Mini-Mental Status Exam; Trail A/Trail B, Trail-Making Forms A & B; FAS, Verbal Fluency F-A-S test; M, male; F, female; SUB, serial subtraction; VF, verbal fluency; NA, not available.

Single-task standing (hypotheses 1/1A)

During the usual single-task standing condition (hard surface, eyes open), there was no difference between the two groups for any sway outcome (t-test p=0.191 – 0.906). Greater postural challenge significantly increased all sway outcomes irrespective of the group (Standing main effect p<0.001) (Table 2). Only the AP amplitude showed a significant Group x Standing interaction (p=0.004), with prosthesis users having greater sway than controls. Similarly, AREA showed a trend towards increased sway in prosthesis users (Group x Standing interaction p=0.055). These results largely confirmed hypotheses 1 and 1A.

Table 2
Mean (SD) and 2-way ANOVA analysis for each sway parameter during single-task standing for prosthesis users (PU) and controls (Ctrl) under the usual and difficult standing conditions. Significant values in bold. The results indicate increased sway with greater postural challenge, largely confirming hypotheses 1 and 1A.

Parameter/ Group		Single-Task Standing		Main Effects		Interaction
		Usual	Difficult	Group	Standing	Group x Standing
PL						
	PU	31.6 (12.9)	117.4 (61.9)	0.215	**<0.001**	0.272
	Ctrl	26.0 (9.0)	90.4 (56.0)			
AREA						
	PU	2.20 (0.90)	16.6 (12.6)	0.061	**<0.001**	0.055
	Ctrl	1.67 (1.09)	9.37 (5.79)			
AP						
	PU	1.1 (0.5)	3.7 (1.5)	**0.005**	**<0.001**	**0.004**
	Ctrl	0.93 (0.4)	2.2 (0.73)			
ML						
	PU	2.4 (0.6)	6.0 (1.1)	0.331	**<0.001**	0.663
	Ctrl	2.5 (1.0)	5.6 (1.5)			

PL, path length (cm); AREA, 95% area (cm^2); AP, anterior-posterior amplitude (cm); ML, medial-lateral amplitude (cm).

Single-task vs. no-prioritization dual-task standing (hypotheses 2)

The prosthesis users showed a significant main effect of Task for all outcomes (p=0.001 – 0.028) and no significant Task x Standing interactions (p=0.133 – 0.714), indicating a greater increase in sway under the dual-task than single-task standing that was not affected by increased postural challenge (Figure 1). Thus hypothesis 2 was rejected, implicating the posture second strategy in prosthesis users.

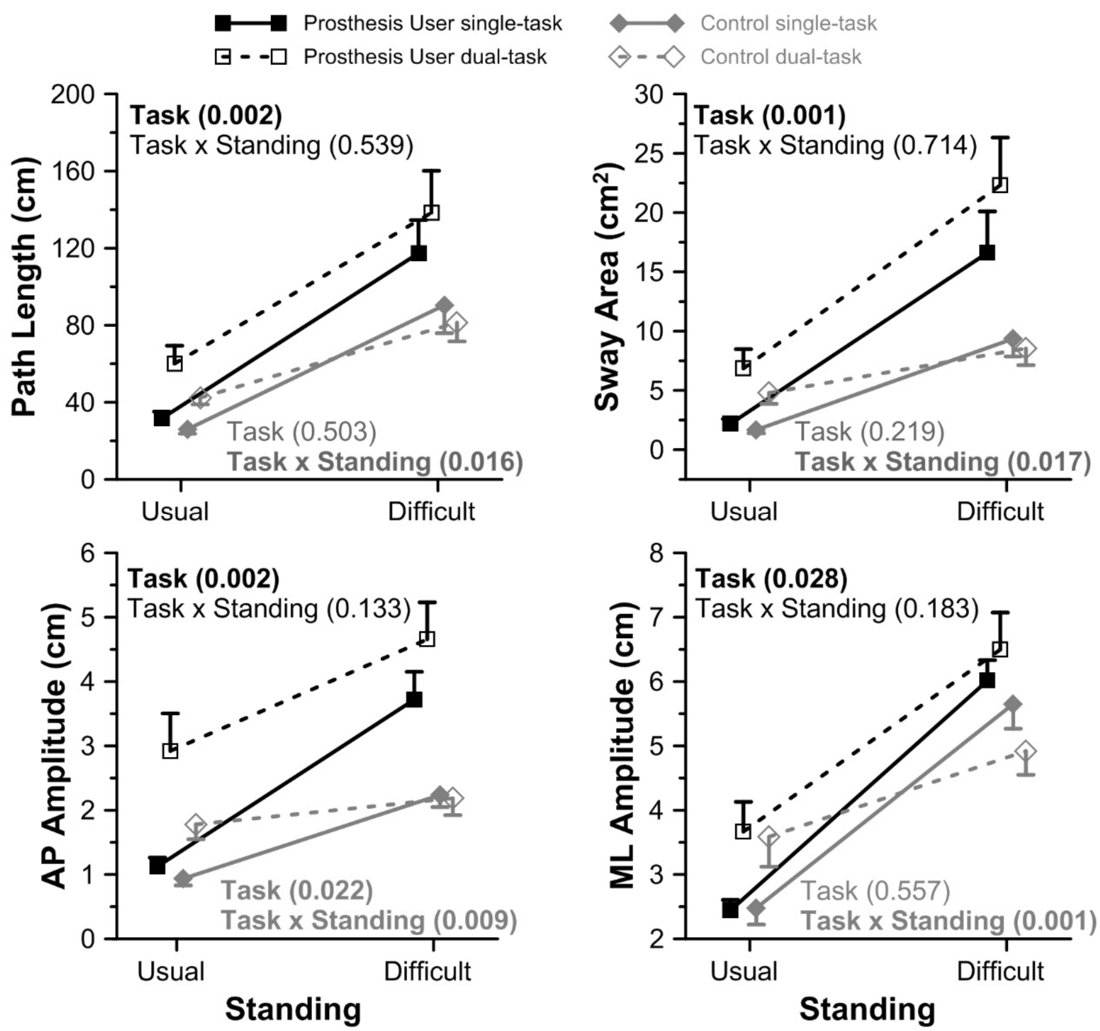

Fig. 1. Singe-task (solid line) vs. no-prioritization dual-task (dashed line) sway parameters between the usual and difficult standing conditions (mean and standard error). Prosthesis users (black squares) significantly increased sway when concurrently performing a cognitive task regardless of standing condition (main effect of Task), supporting the use of the posture second strategy. In controls (gray diamonds), the significant Task x Standing interaction was due to the increase in sway from usual to difficult standing in the single-task but not dual-task condition (note the difference in slopes of the gray solid line compared to the dashed line), which suggests that controls followed the posture first strategy.

The control subjects showed significant Task x Standing interactions for all outcomes (p=0.001 – 0.017). The interactions were due to a *smaller* increase in the dual-task than single-task sway under the difficult standing condition, also evident as the smaller slope from usual to difficult standing condition for the dual-task than single-task standing (Figure 1). These results support hypothesis 2 and implicate the use of the posture first strategy in control subjects. Only the AP amplitude showed a significant effect of Task (p=0.022). The main effect of Standing was significant for all parameters in both groups (p≤0.001).

No-prioritization vs. prioritization sway dual-task cost (hypothesis 3)

The sway dual-task cost for the cognitive no-prioritization and prioritization conditions and the within group statistics are presented in Figure 2. In prosthesis users, the main effect of Instruction was only significant for PL (p=0.030), indicating decreased sway dual-task cost with the instruction to prioritize the cognitive task over the no-prioritization condition. AREA had a significant Instruction x Standing interaction (p=0.041), showing that only with instruction to prioritize the cognitive task did the sway dual-task cost decrease in the difficult standing condition. These results refute hypothesis 3 in prosthesis users, and, therefore, the use of the posture second strategy when asked to prioritize the cognitive task while standing. Only the ML amplitude showed a significant effect of Standing (p=0.049), with less dual-task cost during the difficult standing condition. The AP amplitude followed a similar trend (p=0.078).

In control subjects, the main effect of Instruction was only significant for PL (p=0.028; AREA, AP, ML amplitude p=0.169 – 0.700), indicating less dual-task cost with the instruction to prioritize the cognitive task. These findings also refute hypothesis

3 in controls, which is again inconsistent with the posture second strategy when standing under greater cognitive challenge. Controls subjects further showed a significant main effect of Standing for all sway parameters (p=0.001 – 0.014), indicating less dual-task cost in the difficult than usual standing condition.

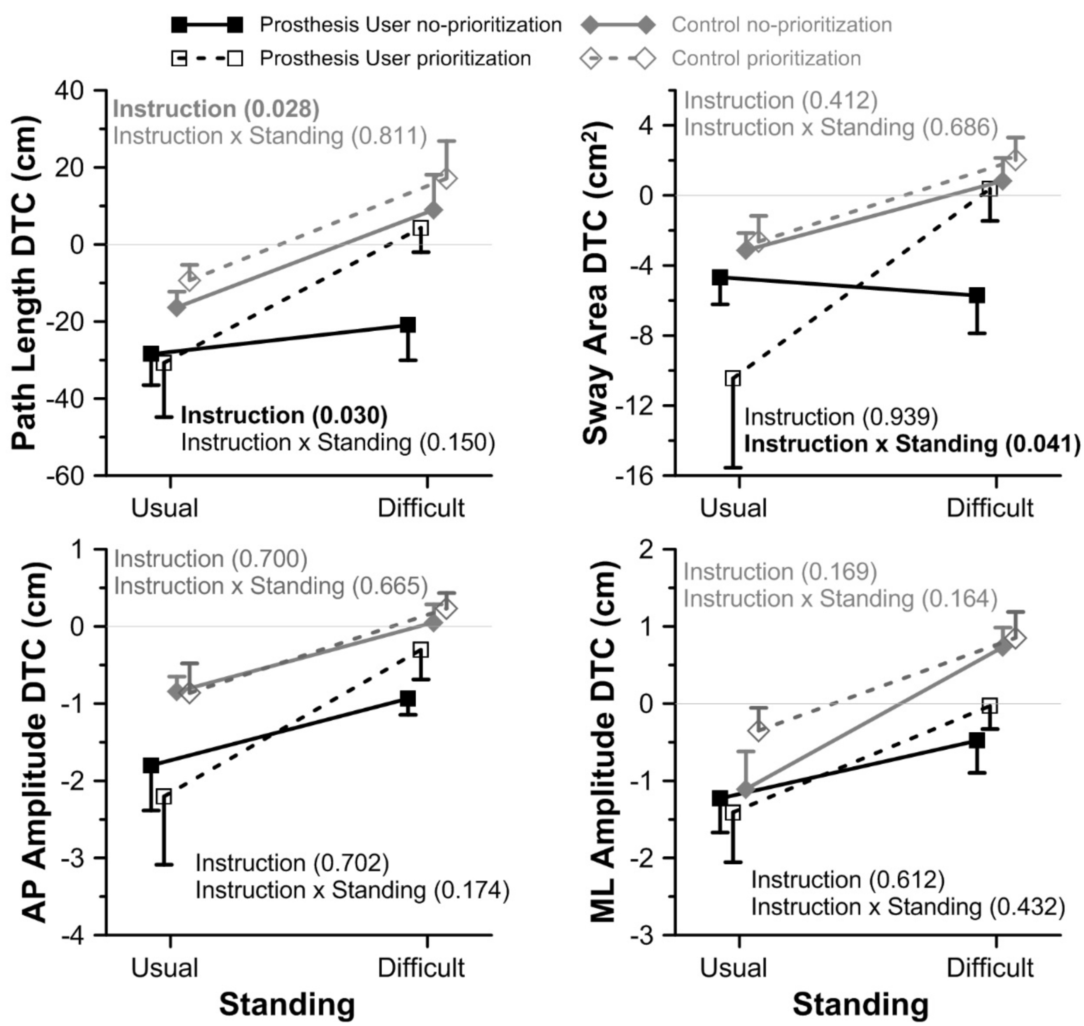

Fig. 2. Dual-task cost (mean and standard error) for each sway parameter for the no-prioritization (solid line) and prioritization (dashed line) conditions. The instruction to prioritize the cognitive task during the difficult standing condition resulted in significantly smaller dual-task cost for path length and 95% sway area in prosthesis users (black square) and for path length in controls (gray diamond), with no changes in the anterior-posterior (AP) or medial-lateral (ML) amplitude. The results confirm hypothesis 3A in both prosthesis users and controls, consistent with posture first strategy.

Cognitive dual-task cost

Across all conditions, the mean cognitive dual-task cost was positive, indicating that on average both groups performed better on the cognitive task while standing. There was a significant Group x Instruction interaction (p=0.026) due to greater improvement in cognitive dual-task performance in prosthesis users than controls when instructed to prioritize the cognitive task (Figure 3). The standing condition (usual vs. difficult) did not have a significant impact on the cognitive dual-task cost (p=0.843).

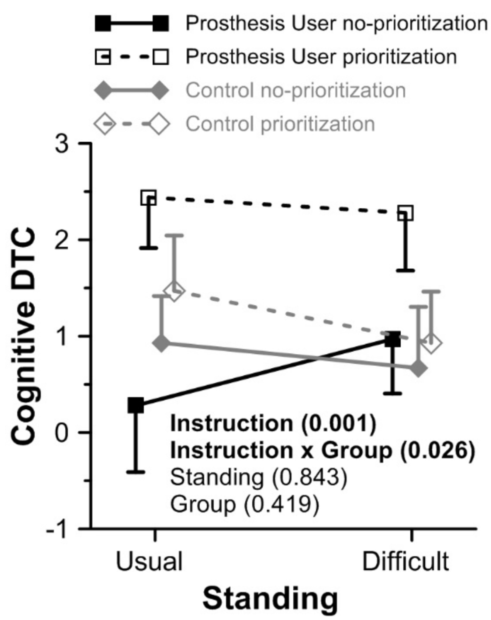

Fig. 3. Dual-task cost (mean and standard error) for the selected cognitive task for the no-prioritization (solid line) and prioritization (dashed line) conditions. The positive cost represents an improvement in performance from seated to standing. Prioritization of the cognitive task only significantly improved performance in prosthesis users (Instruction x Group interaction; black square). The standing condition (Usual vs. Difficult) did not impact performance on the cognitive task in either group.

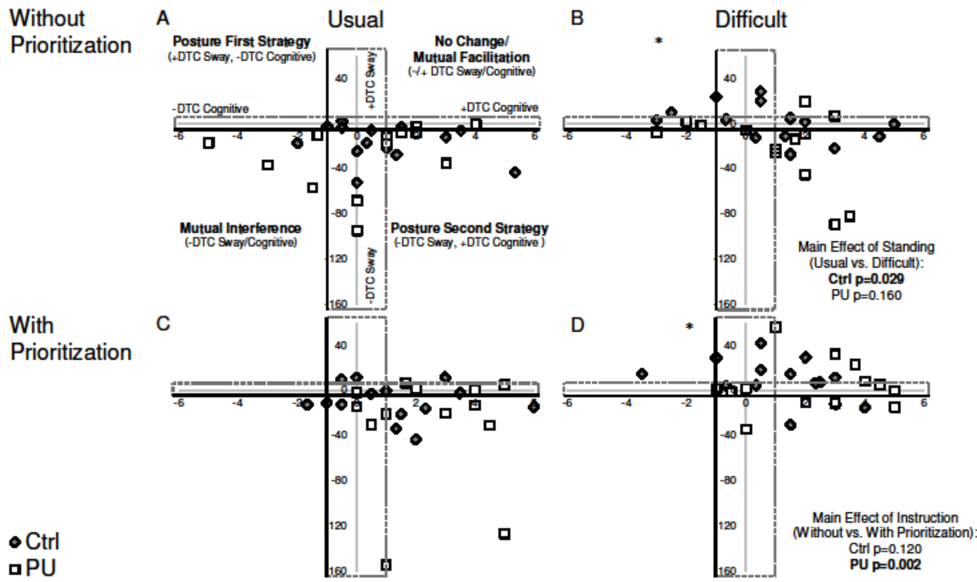

Fig. 4. Posture vs. cognitive dual-task cost for the path length in the usual (A and C) and difficult (B and D) standing conditions for the no-prioritization (A and B) and prioritization (C and D) dual-tasking in controls (gray diamonds) and prosthesis users (white squares). The dashed lines around each axis designate areas of no dual-task cost (path length ± 5.5 cm, cognitive ± 1 point). The thick vertical and horizontal lines in each graph delineate 4 regions consistent with 4 dual-task outcomes: Posture First Strategy (top left), No Change/Mutual Facilitation (top right), Posture Second Strategy (bottom right), and Mutual Interference (bottom left). Note the similar distribution of subjects between the two groups in the usual standing condition with most subjects in the Posture Second region (A and C). The increase in postural challenge (B and D), regardless of the instruction, resulted in a significant proportion of the control subjects moving from the Posture Second region in the usual standing condition to either Posture First or No Change/Mutual Facilitation regions in the difficult standing condition (p=0.029), with no effect on the distribution of prosthesis users (p=0.160). Conversely, the instruction to prioritize the cognitive task (C and D), regardless of standing condition, resulted in a significant shift of the prosthesis users from the Mutual Interference and Posture Second regions in the no-prioritization condition to the No Change/Mutual Facilitation region in the prioritization condition (p=0.002), with no impact on the distribution of controls (p=0.120). The asterisk in each difficult standing condition graph refers to a control data point falling outside the axis range, which was included in the statistical analysis.

Sway vs. cognitive dual-task cost under increasing challenge

To determine how individual subjects in each group changed the strategy under increasing postural and cognitive challenge, we examined frequency distributions between the 4 regions of the plot defined by the sway dual-task cost along the Y-axis and the cognitive dual-task cost along the X-axis (statistics run for PL and AREA, PL is shown in Figure 4). Figure 4A shows that the distribution of prosthesis users and controls during the usual standing without cognitive prioritization was similar, with most subjects in the postural interference/cognitive facilitation (posture second strategy) region (bottom-right quadrant of each plot). As the postural demand increased from the usual to the difficult standing condition, control subjects moved from the posture second strategy or mutual interference regions into the posture first strategy (postural facilitation/cognitive interference) or mutual facilitation regions (main effect of Standing PL p=0.029; AREA p=0.036). This can be seen by comparing the distribution of controls between Figure 4A/B and Figure 4C/D. The instruction to prioritize the cognitive task did not affect controls (Instruction main effect p>0.1). Conversely, increased postural challenge did not affect prosthesis users (Standing main effect p>0.16), but the instruction to prioritize the cognitive task resulted in their move from the posture second strategy or mutual interference regions to the mutual facilitation region (Instruction main effect PL p=0.002; AREA p=0.032) (compare Figure 4A/C vs. 4B/D). The interaction between Standing and Instruction was not significant for PL or AREA in either group. These findings substantiate the results of group analyses.

Discussion

This study provides several novel findings specifically related to the selection and adaptation of postural strategy under increasing postural and cognitive challenge. The results confirmed that prosthesis users can maintain stability in a manner similar to non-amputee controls when standing on a hard surface with eyes open. With an increase in the postural challenge (soft surface/eyes closed), sway increased similarly in both groups. In controls, adding a cognitive task without specific instruction on prioritization resulted in a smaller increase in sway at the greater postural challenge in comparison to standing only, suggesting the use of the posture first strategy. In contrast, prosthesis users under the same conditions further increased sway, consistent with the posture second strategy, which was supported by the improved cognitive performance. Finally, the specific instruction to focus on the cognitive task improved cognitive performance without negatively impacting sway in either group, indicating a more complex cognitive-motor interaction beyond just posture prioritization.

No difference between the prosthesis users and controls while standing with eyes open on a hard surface confirmed our first hypothesis, suggesting that the recruited prosthesis users had sufficient resources in the postural reserve to maintain stability during usual standing. While contradicting reports of increased sway in prosthesis users during usual standing [15, 25-27], this is in agreement with findings that more challenging standing conditions may be required to observe differences in postural stability between prosthesis users and controls [16, 17]. Differences in age, health status, or the cause of amputation, as well as improvements in prosthetic componentry, may explain discrepancies compared to some older studies.

With increased postural challenge (softer standing surface, no visual input), the two primary outcomes (PL, AREA) and one secondary outcome (ML amplitude) increased in both prosthesis users and controls, in line with the first part of hypothesis 1A. However, the second part of hypothesis 1A was partially supported because only the AP amplitude increased more in prosthesis users compared to controls with a greater postural challenge. The significant increase in the AP amplitude was likely behind the near significant increase in AREA. Since motion in the AP direction is primarily driven by ankle activation [28], it comes as no surprise that AP movement revealed greater sway in prosthesis users since the prosthetic device cannot fully restore postural control at the ankle. Greater impairment in the AP direction among active prosthesis users has been previously reported [25].

Both prosthesis users and controls increased sway while standing and concurrently performing a cognitive task without prioritization instruction; however, the increase in sway with greater postural challenge was smaller in controls, supporting the engagement of a strategy that focuses on maintaining balance, such as the posture first strategy (hypothesis 2). In contrast, prosthesis users did not limit the increase in sway with greater postural challenge, following the posture second strategy. The employment of the posture second strategy with increased postural challenge by prosthesis users is contrary to our initial assumptions and suggests that they are willing to allocate resources in a way that leads to increased risk to stability. This behavior in prosthesis users may pose a real threat since it has also been reported in other groups with sensorimotor impairments prone to falls [3, 5, 8, 9, 11], and greater dual-task cost has been associated with increased fall risk [29-31]. While the propensity for posture second strategy in prosthesis users seems at first unexpected, such behavior is consistent with our previous report on lower limb preference among 11 goal-oriented tasks [32]. We found that prosthesis users do not always rely on the intact leg for support and the prosthetic leg for executing goal-oriented lower limb tasks (e.g., stepping on a bug, hitting a moving target, kicking a ball), as expected with the posture first strategy. Instead, we frequently found the opposite, where the prosthetic leg was used for support during a variety of goal-oriented tasks, which generally follows the posture second strategy. The results of the two studies support the notion that the selection of postural strategy depends on the level of perceived threat to balance weighted against individual goals, which accomplishment is likely also influenced by their relevance and set priorities.

Further challenge imposed by the instruction to improve the performance on the cognitive task led to no change or reduced dual-task cost for the majority of sway parameters in both controls and prosthesis users, contrary to hypothesis 3 and particularly in the difficult standing condition. This suggests that the posture second strategy was not employed in controls and implies a shift away from the posture second in prosthesis users.

The cognitive dual-task cost was on average positive in both prosthesis users and controls across all conditions (Figures 3 and 4). The findings that prosthesis users showed a negative dual-task cost for sway measures during increasingly more difficult standing without prioritization (Figure 1) suggest that, under these conditions, they employed the posture second strategy, which was not the case in controls. The positive cognitive dual-task cost (improved performance) along with sway that was either improved or not significantly changed in controls under all conditions (Figures 1 and 2), and in prosthesis users during standing under cognitive prioritization (Figure 2), suggests mutual facilitation rather than the posture first strategy. The significant Group x Instruction interaction indicates the prosthesis users moved away from the posture second strategy while standing under greater cognitive demand, without further change in controls.

These group results are also supported after examining the behavior of individual subjects through the combined analysis of sway and cognitive dual-task costs (Figure 4). When instructed to prioritize the cognitive task, 3 observations became more prevalent in the prosthesis users; improved cognitive performance over the seated-baseline without increasing sway (note rare negative scores for cognitive dual-task cost in Figure 4C/D), positive sway dual-task cost (lesser sway) in the difficult standing condition (Figure 4D), and departure from the posture second strategy or mutual interference regions (compare Figure 4A/C and 4B/D). With greater postural challenge, and regardless of the prioritization instruction, controls also moved out of the posture second strategy region occupied in the usual standing condition. However, when considering both the sway and cognitive dual-task cost, it became apparent that neither group primarily engaged the posture first strategy. Instead, most subjects migrated into the No Impact/Mutual Facilitation regions (Figure 4B/D, top right section). While the posture first strategy implies that the prioritization of posture should come at the expense of a concurrent task, most of our subjects either maintained or improved performance on the cognitive task while simultaneously improving (reducing) postural sway.

Several theories have been proposed to explain the neural basis of dual-task performance. The central capacity-sharing theory [33, 34] and the similar cross-domain completion model [2] propose that if simultaneous tasks compete for the same resources, one or both tasks will have a decrease in performance. While this fits well with our initial assumptions of a trade-off in performance between the two tasks primarily driven by postural difficulty, these two theories do not allow for simultaneous improvements in performance [2], as we observed during the more challenging conditions (Figure 1 B/D).

The bottleneck theory suggests that if the same neural networks are used to process the concurrent information, performance declines as the networks are overloaded forming a bottleneck [35]. Our findings of greater dual-task impact during easier conditions, but smaller dual-task impact during more challenging standing or cognitive condition, suggest two possibilities. First, the same neural networks were not used for the task completion and the initial dual-task impact was due to factors other than a bottleneck. Alternatively, the networks were initially shared, resulting in a bottleneck in the cases of higher dual-task cost, but increased challenge led to switching to other networks. However, the concurrent increase in cognitive performance and a decrease in sway (i.e., cognitive-motor facilitation), as reported here, is best explained by the level of alertness hypothesis, which suggests that with increased demand more resources get recruited [6, 36]. Such resources seem readily available as demonstrated by improved dual-task performance after increasing neural activation by anodal transcranial direct current stimulation [37].

There are several broader implications of our results obtained in prosthesis users. Considering general agreement with previous studies in brain disorders [7, 9, 12], our results lend support for a unifying view that individuals with sensorimotor impairments, whether caused by a partial leg loss or neurological damage, adopt strategies that allow achieving desired goals as successfully as possible even if it comes with certain risks. This may well be an adjustment to go on with their lives as usual, the success of which would depend on availability and capacity to engage the remaining cognitive-motor resources. Greater challenge, however, requires greater involvement of resources, which, if available, may improve performance. In the case of prosthesis users, explicit instruction to improve performance on the cognitive task likely resulted in recruitment of global resources improving not only overt (cognitive task) but also covert (posture control) actions. This fits well with brain imaging findings of increased and complex cognitive activation while dual-tasking, suggesting that concurrent performance of challenging tasks may activate overlapping cognitive and motor regions [38]. While selecting at first the posture second strategy and later accessing additional resources as needed is feasible in individuals with preserved brain functions, as demonstrated in prosthesis users here, the same may not be a viable strategy for those with central nervous system disorders, depending on their nature and severity.

Limitations

This study has some limitations. As the first investigation into posture strategy selection in prosthesis users, the sample was not homogeneous with respect to age, the cause of amputation, or type of componentry. The mean age of our prosthesis users was lower than the average for this population [39] and most were active community ambulators, which limits generalization of findings. Although older adults show increased dual-task cost [40], it remains unknown if the compounded effect of age in prosthesis users would cause greater dual-task impact, trigger increased alertness earlier, or invoke alternative strategies. While in this small sample the results did not differ between subjects with traumatic and vascular amputations, widespread vascular disease may lead to cognitive impairments and impact dual-task performance [41]. However, the baseline cognitive performance was no different between these groups and mostly active prosthesis users participated in the study, which may have reduced any such differences. The use of different prosthetic components may have altered postural control and biased the results. Thus, further work should evaluate the impact of age, the cause of amputation, and componentry on dual-task performance in this population. Also, this study only examined below-knee prosthesis users; although a more proximal amputation is considered to cause greater sensorimotor impairment, performing a cognitive task did not alter walking in above-knee prosthesis users when instructed to focus on cognitive task performance [19]. Based on our results, more impaired prosthesis users may engage increased alertness sooner to limit the impact of dual-tasking on dynamic stability, which warrants further studies. The condition with prioritization of the standing task was not collected, as pilot testing revealed frequent confusion with this instruction. The inherent simplicity of standing made it difficult for our subjects to comply with the intended goal of such instruction. Only a single dual-task trial was examined for each standing condition, which may give a limited picture of actual behavior. However, mental and physical fatigue from repeated trials was considered a greater threat to validity. Since the prioritization conditions were not randomized, some contribution of a learning effect was possible. Thus, we evaluated each aim for within-group differences in task order and none were significant for either the standing task ($p \geq 0.09$) or cognitive task ($p \geq 0.2$), effectively minimizing this concern.

Conclusion

Both prosthesis users and controls allow for greater sway while concurrently performing a cognitive task under less demanding postural condition (hard surface, eyes open), suggesting the posture second strategy. However, performing a cognitive task under increased postural challenge (soft surface, eyes closed) leads to the shift from the posture second toward the posture first strategy in non-amputee controls but retention of the posture second strategy in below-knee prosthesis users. Since the posture second strategy implies greater unsteadiness, prosthesis users appear exposed to a greater risk of fall when performing multiple tasks, perhaps to maintain activity level and lifestyle similar to their non-amputee peers. However, when faced with highly demanding cognitive and postural tasks, both prosthesis users and controls engage additional resources resulting in better cognitive and motor performance, which may be explained by increased alertness rather than posture prioritization alone.

References

[1] Woollacott M, Shumway-Cook A. Attention and the control of posture and gait: a review of an emerging area of research. Gait Posture. 2002; 16:1-14.

[2] Lacour M, Bernard-Demanze L, Dumitrescu M. Posture control, aging, and attention resources: Models and posture-analysis methods. Clin Neurophysiol. 2008; 38:411-21.

[3] Yogev-Seligmann G, Hausdorff JM, Giladi N. Do we always prioritize balance when walking? Towards an integrated model of task prioritization. Mov Disord. 2012; 27:765-70.

[4] Bronstein AM, Brandt T, Woollacott MH, Nutt JG. Clinical Disorders of Balance, Posture and Gait. 2nd ed. London: Arnold Publishers; 2004.

[5] Bloem BR, Grimbergen YA, van Dijk JG, Munneke M. The "posture second" strategy: a review of wrong priorities in Parkinson's disease. J Neurol Sci. 2006; 248:196-204.

[6] Bisson EJ, Lajoie Y, Bilodeau M. The influence of age and surface compliance on changes in postural control and attention due to ankle neuromuscular fatigue. Exp Brain Res. 2014; 232:837-45.

[7] Yogev-Seligmann G, Hausdorff JM, Giladi N. The role of executive function and attention in gait. Mov Disord. 2008; 23:329-42.

[8] Shumway-Cook A, Woollacott M, Kerns KA, Baldwin M. The effects of two types of cognitive tasks on postural stability in older adults with and without a history of falls. J Gerontol A Biol Sci Med Sci. 1997; 52:232-40.

[9] Yogev-Seligmann G, Rotem-Galili Y, Dickstein R, Giladi N, Hausdorff JM. Effects of explicit prioritization on dual task walking in patients with Parkinson's disease. Gait Posture. 2012; 35:641-6.

[10] Kelly VE, Eusterbrock AJ, Shumway-Cook A. Factors influencing dynamic prioritization during dual-task walking in healthy young adults. Gait Posture. 2013; 37:131-4.

[11] Swanenburg J, de Bruin ED, Favero K, Uebelhart D, Mulder T. The reliability of postural balance measures in single and dual tasking in elderly fallers and non-fallers. BMC Musculoskelet Disord. 2008; 9:162.

[12] Plummer-D'Amato P, Altmann LJP, Saracino D, Fox E, Behrman AL, Marsiske M. Interactions between cognitive tasks and gait after stroke: A dual task study. Gait Posture. 2008; 27:683-8.

[13] Miller WC, Speechley M, Deathe AB. The prevalence and risk factors of falling and fear of falling among lower extremity amputees. Arch Phys Med Rehabil. 2001; 82:1031-7.

[14] Hunter SW, Batchelor F, Hill KD, Hill A-M, Mackintosh S, Payne M. Risk Factors for Falls in People With a Lower Limb Amputation: A Systematic Review. PMR. 2017; 9:170-80.

[15] Geurts AC, Mulder TW, Nienhuis B, Rijken RA. Postural reorganization following lower limb amputation. Possible motor and sensory determinants of recovery. Scand J Rehabil Med. 1992; 24:83-90.

[16] Mohieldin A, Chidambaram A, Sabapathivinayagam R, Al Busairi W. Quantitative assessment of postural stability and balance between persons with lower limb amputation and normal subjects by using dynamic posturography. Maced J Med Sci. 2010; 3:138-43.

[17] Vanicek N, Strike S, McNaughton L, Polman R. Postural responses to dynamic perturbations in amputee fallers versus nonfallers: a comparative study with able-bodied subjects. Arch Phys Med Rehabil. 2009; 90:1018-25.

[18] Vrieling AH, van Keeken HG, Schoppen T, Otten E, Hof AL, Halbertsma JP, et al. Balance control on a moving platform in unilateral lower limb amputees. Gait Posture. 2008; 28:222-8.

[19] Morgan SJ, Hafner BJ, Kelly VE. The effects of a concurrent task on walking in persons with transfemoral amputation compared to persons without limb loss. Prosthet Orthot Int. 2016; 40:490-6.

[20] Howard CL, Wallace C, Abbas J, Stokic DS. Residual standard deviation: Validation of a new measure of dual-task cost in below-knee prosthesis users. Gait Posture. 2017; 51:91-6.

[21] Geurts AC, Mulder TH. Attention demands in balance recovery following lower limb amputation. J Mot Behav. 1994; 26:162-70.

[22] Plummer P, Eskes G. Measuring treatment effects on dual-task performance: a framework for research and clinical practice. Front Hum Neurosci. 2015; 9:225.

[23] Yogev-Seligmann G, Rotem-Galili Y, Mirelman A, Dickstein R, Giladi N, Hausdorff JM. How does explicit prioritization alter walking during dual-task performance? Effects of age and sex on gait speed and variability. Phys Ther. 2010; 90:177-86.

[24] Borkowski JG, Benton AL, Spreen O. Word fluency and brain damage. Neuropsychologia. 1967; 5:135-40.

[25] Buckley JG, O'Driscoll D, Bennett SJ. Postural sway and active balance performance in highly active lower-limb amputees. Am J Phys Med Rehabil. 2002; 81:13-20.

[26] Isakov E, Mizrahi J, Ring H, Susak Z, Hakim N. Standing sway and weight-bearing distribution in people with below-knee amputations. Arch Phys Med Rehabil. 1992; 73:174-8.

[27] Hermodsson Y, Ekdahl C, Persson B, Roxendal G. Standing balance in trans-tibial amputees following vascular disease or trauma: a comparative study with healthy subjects. Prosthet Orthot Int. 1994; 18:150-8.

[28] Winter DA. Human balance and posture control during standing and walking. Gait Posture. 1995; 3:193-214.

[29] Sample RB, Jackson K, Kinney AL, Diestelkamp WS, Reinert SS, Bigelow KE. Manual and Cognitive Dual Tasks Contribute to Fall-Risk Differentiation in Posturography Measures. J Appl Biomech. 2016; 32:541-7.

[30] Etemadi Y. Dual task cost of cognition is related to fall risk in patients with multiple sclerosis: a prospective study. Clin Rehabil. 2017; 31:278-84.

[31] Muir-Hunter SW, Wittwer JE. Dual-task testing to predict falls in community-dwelling older adults: a systematic review. Physiotherapy. 2016; 102:29-40.

[32] Howard C, Wallace C, Stokic DS. Lower limb preference on goal-oriented tasks in unilateral prosthesis users. Gait Posture. 2012; 36:249-53.

[33] Beurskens R, Steinberg F, Antoniewicz F, Wolff W, Granacher U. Neural correlates of dual-task walking: Effects of cognitive versus motor interference in young adults. Neural Plast. 2016; 2016:1-9.

[34] Tombu M, Jolicoeur P. A central capacity sharing model of dual-task performance. J Exp Psychol Hum Percept Perform. 2003; 29:3-18.

[35] Fujita H, Kasubuchi K, Wakata S, Hiyamizu M, Morioka S. Role of the Frontal Cortex in Standing Postural Sway Tasks While Dual-Tasking: A Functional Near-Infrared Spectroscopy Study Examining Working Memory Capacity. Biomed Res Int. 2016; Epub: Feb 3; 10.1155/2016/7053867.

[36] Bonnet CT, Baudry S. Active vision task and postural control in healthy, young adults: Synergy and probably not duality. Gait Posture. 2016; 48:57-63.

[37] Wrightson JG, Twomey R, Ross EZ, Smeeton NJ. The effect of transcranial direct current stimulation on task processing and prioritisation during dual-task gait. Exp Brain Res. 2015; 233:1575-83.

[38] Metzger FG, Ehlis AC, Haeussinger FB, Schneeweiss P, Hudak J, Fallgatter AJ, et al. Functional brain imaging of walking while talking - An fNIRS study. Neuroscience. 2017; 343:85-93.

[39] NLLIC. Diabetes and lower extremity amputations. In: America ACo, editor. http://www.amputee-coalition.org/fact_sheets/diabetes_leamp.html: National Limb Loss Information Center; 2008.

[40] Liston MB, Bergmann JH, Keating N, Green DA, Pavlou M. Postural prioritization is differentially altered in healthy older compared to younger adults during visual and auditory coded spatial multitasking. Gait Posture. 2014; 39:198-204.

[41] Coffey L, O'Keeffe F, Gallagher P, Desmond D, Lombard-Vance R. Cognitive functioning in persons with lower limb amputations: a review. Disabil Rehabil. 2012; 34:1950-64.

CHAPTER 6

THE IMPACTS OF POSTURAL AND COGNITIVE CHALLENGES ON THE SPECTRAL COMPONENTS OF SWAY IN PROSTHESIS USERS AND CONTROL SUBJECTS

Abstract

Lower-limb amputation impairs postural control capabilities, however, little is known about the specific impact amputation has on the roles of the sensory systems involved. Spectral analysis of the center of pressure signal while standing has identified frequency bands associated with postural adjustments driven by the visual, vestibular, and somatosensory systems. Using wavelet analysis, the spectral features of the center of pressure signal in 13 below-knee prosthesis users and 14 control subjects were characterized in the medial-lateral (ML) and anterior-posterior (AP) directions. Subjects were tested in the baseline condition (eyes opened, hard surface), with eyes closed, while standing on a soft surface, and while performing a cognitive task (dual-tasking). During single-task standing, the more difficult standing conditions increased the total spectral power in both groups ($p \le 0.005$); however, the increase was greater in the AP direction in prosthesis users ($p \le 0.036$). The eyes closed conditions reduced the contribution from the frequency band associated with vision ($p \le 0.005$) and the soft surface condition increased the contribution from the band associated with somatosensation ($p \le 0.03$). Dual-tasking increased the total spectral power in prosthesis users more than in control subjects ($p \le 0.05$) and reduced the ML contribution from the frequency band associated with vision in both groups ($p<0.001$). Prosthesis users also had a smaller ML contribution from the somatosensory band on the prosthetic than the intact side ($p<0.001$). Results

demonstrate that postural control is more disrupted by dual-tasking in prosthesis users than in control subjects and that prosthesis users rely more heavily on somatosensory feedback from the intact side than from the prosthetic side. These results support the use of spectral analysis to evaluate the contributions from sensory systems involved in postural control and suggest that postural control in prosthesis users may be improved by reducing the attentional demands or by supplementing somatosensory feedback on the prosthetic side.

Introduction

The visual, vestibular, and somatosensory systems are known contributors to postural control [1, 2]. In our previous analysis of sway in below-knee prosthesis users we evaluated how increasing postural challenge by eliminating or limiting input from a sensory system and adding a cognitive challenge (dual-task) impacted temporal-spatial sway characteristics and the relation to postural control strategies [3]. While the study provided insight into general changes in postural control, the level of analysis did not allow for investigation of the impact on the contributions from the specific sensory systems.

Studies in spectral analysis of the center of pressure (CoP) signal during postural control have identified frequency bands that are associated with postural adjustments directed from each sensory system. These studies, which have evaluated populations with impaired or absent sensory systems [4-6] or utilized experimental manipulations [7-16], have identified consistent trends in spectral power contributions. Collectively, these studies have produced a general consensus on the frequency bands and the associated sensory systems: vision is associated with very low frequencies (< 0.1 Hz), vestibular low frequencies (~ 0.1-0.5 Hz), somatosensory middle frequencies (~ 0.5-1 Hz), and feedforward or open-loop control is associated with high frequencies (> 1 Hz). A more thorough summary of these reports in the literature has been provided in the background chapter (Chapter 2).

Spectral analysis of the CoP signal has also been found to be more sensitive than traditional measures in detecting the effects of standing conditions or differentiating between groups [8, 10, 17]. Dual-tasking has been found to increase spectral power even when changes in temporal-spatial indices were not present [8, 10]. Higher spectral power may represent more rapid changes in the CoP trajectory [8]. More CoP movement, measured through traditional or spectral analysis, is often associated with lower stability [10, 18-20]. Since spectral analysis may be more sensitive to the effects of dual-tasking, evaluation of the spectral power of the CoP signal may be useful in further characterizing the impact of dual-tasking on posture control capabilities that has been observed in lower-limb amputees [3].

Wavelet analysis has been suggested to be superior to traditional Fourier transforms for analysis of CoP signals that seeks to characterize spectral power in specific frequency bands [2, 8, 9, 21-23]. Wavelet analysis uses variable sized, time-scale specific windows to deconstruct the signal into time-scale bands; the time-scales can then be transformed to frequencies. Wavelet analysis enables more accurate deconstruction of time varying, non-stationary signals, such as posture CoP signals [8, 9, 23] and is better at characterizing the spectral power in non-dominant frequencies [8, 9]. This is particularly useful when evaluating changes in multiple frequency bands in the CoP signal, which is typically dominated by energy in the very low frequency band [8, 9].

Finally, there is evidence that wavelet analysis performs better than traditional techniques on short-duration time series signals [8]. This feature can be particularly important for evaluating postural control in impaired populations or difficult standing conditions when longer collection times are impractical [8].

Lower-limb amputation alters both efferent and afferent control of posture due to missing joints, musculature, and altered sensory input. It is unknown how this disruption impacts the contribution of somatosensory or other sensory systems directing postural control or if it affects the ability to adapt to more difficult postural control conditions. In our previous work we identified that during a difficult standing condition in which subjects stood with eyes closed on a soft surface, the anterior-posterior (AP) sway amplitude, but not the medial-lateral (ML), increased more in lower-limb amputees than in control subjects. However, while performing a cognitive task during the difficult standing condition (dual-tasking) prosthesis users increased both AP and ML amplitudes, while control subjects did not. Since disruptions to the visual and somatosensory systems were used to create the difficult standing conditions and prosthesis users exhibited differential performance from control subjects we sought to characterize the contributions of each sensory system to postural control in each group.

The overall objective of this follow-up study was to determine if spectral analysis can identify changes in the frequency profile induced by subjecting prosthesis users and control subjects to various standing and cognitive load conditions. Our first aim was to evaluate differences between groups and changes across standing conditions during single-task standing. Based on our previous results, we proposed that during single-task standing prosthesis users will have a greater increase than control subjects in total spectral power in the AP direction (hypothesis 1). We also predicted that the more difficult standing conditions would alter the distribution of spectral power across the frequency bands associated with the visual, vestibular, somatosensory systems, and open-loop control. Specifically, we hypothesized that the eyes closed conditions would reduce the relative contribution from the very low (vision) frequency band as other systems take on a greater role (hypothesis 2A) and the soft surface would increase demand on the middle (somatosensory) frequency band as the system works harder to compensate for the reduced feedback (hypothesis 2B). In our second aim, we evaluated the impact of increased cognitive load from dual-tasking. We hypothesized that dual-tasking would impact the prosthesis users but not control subjects (hypothesis 3). Theses hypotheses were tested by evaluating the impact of the various standing and cognitive load conditions on total spectral power and on the relative spectral power from each frequency band.

Since the study focused on unilateral prosthesis users who have an inherent asymmetry in bilateral lower-limb tasks, we also examined differences in the spectral power between each side (prosthetic/intact; dominant/non-dominant) to assess how each side contributes to postural control.

Methods

Participants

A convenience sample of unilateral below-knee prosthesis users (n=16) was recruited by a certified prosthetist. The inclusion criteria were ≥ 1 year since amputation; age 18–80 years; comfortable socket fit; no known balance, neurological, or other health problems that limit daily activities; and able to safely stand without use of an assistive device. Age- and education-matched non-amputee control subjects were recruited from the community (n=17). The study was approved by the institutional review board for human research and all subjects signed the informed consent.

Protocol

The groups' temporal-spatial sway characteristics (path length, area, and medial-lateral and anterior-posterior amplitudes) were previously analyzed and have been reported in [3] and the same data set was used to evaluate the spectral characteristics of sway. Since the data collection and process to obtain the CoP data have been described in detail in [3], only a brief overview is presented here.

Subjects' general cognitive functions were assessed using the Modified Mini-Mental Status Exam (3MS) and processing speed and executive function were assessed using Trail-Making (Trail) forms A and B (scaled T-scores adjusted for race, age, gender, and education; higher scores represent better performance). Demographic and prosthesis users' clinical data were collected through an interview and from medical records.

All standing tasks were performed at shoulder width stance using 2 force plates (Type 4060, 40x60 cm^2, Bertec Corp, Columbus, OH). Force plate data were collected using a Cortex data acquisition system (Motion Analysis Corp., Santa Rosa, CA, sample rate 1,200 Hz, 12-bit analog-to-digital resolution).

Two vision conditions (eyes open and closed) and 2 standing surfaces (hard and soft) were used to create varying levels of postural challenge. The soft standing condition required subjects to stand with each foot on a foam pad (Airex Balance Pad, Sins, Switzerland, 40x50x6 cm^3, 0.726 kg, density 61 kg/m^3).

Subjects performed two 30-s trials for each surface/vision combination during single-task standing; the average of the two trials were used for analysis. Two cognitive tasks were selected for the dual-task paradigm; serial subtraction by 7 from a 3-digit number and a verbal fluency task (listing words starting with a specific letter). Each task was practiced while seated to ensure comprehension. Subjects performed one 30-s trial for each task. Subjects were given no instruction on task prioritization. All single-task standing conditions were collected prior to the dual-task conditions.

Due to variations in personal aptitude, the relative magnitudes of the interfering effect of the two tasks may vary across subjects. The more stressful task identified and analyzed in [3] was utilized for the spectral analysis. The same approach proved to be effective in our prior work [24].

Data processing

Ground reaction forces and moments were used to determine the CoP for each foot and their resultant in the ML and AP directions. The individual side and resultant CoP locations were then filtered with a 4th order low-pass Butterworth filter (cutoff frequency 10 Hz) and down-sampled to 400 Hz to create desired time scale/frequency band resolution while maintaining as much of the signal as possible and to be consistent with procedures reported in the literature [8-10]. Due to sporadic artifacts at the beginning or end of some trials, the middle 5,120 samples (25.6-s of the 30-s trial) were analyzed for consistency between subjects.

The spectral characteristics of the ML and AP CoP signals were evaluated using a discrete wavelet packet decomposition. The mother wavelet Daubechies 4 was used for this analysis as it meets the criteria for this analysis, provided adequate resolution of the signal to the desired bands, and the family has been found useful for CoP analysis in other studies as the structure resembles the shape of the CoP signal [7, 9, 19, 25]. Nine levels were used to decompose the signal, as this provided the longest scale while avoiding edge effects [23].

The total spectral power of the signal in each standing condition below 10.15 Hz and the relative spectral power (%) in the very low [0 – 0.19 Hz), low [0.19 – 0.39 Hz), middle [0.39 – 1.17 Hz), and high [1.17 – 10.15 Hz) frequency bands were calculated. These bands approximate the spectral regions associated with the visual, vestibular, and somatosensory systems and open-loop control, respectively [2, 4-15]. Total spectral power along with the very low and middle frequency bands were considered the primary outcomes for hypothesis testing, the remaining frequency bands were used as secondary measures. All computations were completed in MATLAB® (Mathworks Inc., Natick, MA) utilizing the Wavelet Toolbox and a custom program.

Only subjects analyzed in [3] were included in the analysis (13 prosthesis users and 15 control subjects). One additional control subject was identified as an outlier and exclude from the analysis as the total spectral power was consistently 3 SDs above the group mean despite normal values in the previous analysis.

Statistical analysis

Baseline cognitive performance was compared between prosthesis users and control subjects (unpaired t-test, $p \leq 0.05$). Cognitive function was not used as a covariate as the two groups did not differ on standard tests of cognitive function (3MS, Trails A/B, FAS). The difference in total spectral power between the two groups under the usual single-task standing condition, was tested using the single-task eyes open/hard surface condition (unpaired t-tests, $p \leq 0.05$). To test whether increasing the postural challenge resulted in higher total spectral power in prosthesis users (hypothesis 1), a 2x2x2 mixed ANOVA was used with Group (prosthesis users, control subjects) as the between-subjects factor and Vision (eyes open, eyes closed) and Surface (hard, soft) as the within-subjects factors. The differential response of the two groups was determined by a Group interaction or the main effect of Group ($p \leq 0.05$). The impact of the standing conditions on the relative spectral power from each frequency band were also tested with the same 2x2x2 mixed ANOVA model for each band (hypotheses 2A/B). Since relative spectral power in two frequency bands were used as primary outcome measures, a Bonferroni correction ($p \leq 0.025$) was used for the primary and secondary outcomes.

The effect of the cognitive task on total spectral power and relative spectral power (hypothesis 3) was tested using a 2x2x2x2 mixed ANOVA with Group (prosthesis users, control subjects) as the between-subjects factor and Vision (eyes open, eyes closed), Surface (hard, soft), and Task (single, dual) as the within-subjects factors. The effect of the cognitive task was determined with the main effect of Task or a Task x standing condition (Vision or Surface) interaction ($p \leq 0.05$ total spectral power; $p \leq 0.025$ relative spectral power). The differential response to the cognitive task was determined by a Task x Group interaction ($p \leq 0.05$ total spectral power; $p \leq 0.025$ relative spectral power).

The difference in total spectral power and relative spectral power between the prosthetic/intact or dominant/non-dominant sides was evaluated with a 2x2x2x2 repeated measure ANOVA with Vision (eyes open, eyes closed), Surface (hard, soft), Task (single, dual), and Side (prosthetic/intact or dominant/non-dominant) as factors in each group. A main effect of Side or interaction including Side was used to determine if there was a difference in each side ($p \leq 0.05$ total spectral power; $p \leq 0.025$ relative spectral power). IBM SPSS Statistics 23 (IBM Corp., Armonk, NY) was used for statistical analysis.

Results

The studied sample included 14 control subjects (mean ± SD age 47±17 years, 15±2 years of education, BMI 30±7 kg/m^2, 6 (43%) men) and 13 below-knee prosthesis users (age 46±11 years, 14±3 years of education, BMI 31.4±6 kg/m^2, 9 (70%) men). The amputation occurred 8±7 years earlier (range 1 to 22 years) due to trauma (n=6), infection (n=2), or vascular disease (n=5). All below-knee prosthesis users were rated K3 on the Medicare scale and none used an assistive device. They all used an energy storage and return style foot; the prosthesis for 3 of the subjects also included a hydraulic ankle. Two used a passive suction suspension system, 7 elevated vacuum, and 4 a pin locking system. Most (9 of 13) reported living an active to very active lifestyle.

The baseline cognitive performance did not significantly differ between prosthesis users and control subjects (mean ± SD 3MS 95±5 vs. 96±4; Trail A 48±10 vs. 45±12; Trail B 48±13 vs. 48±12, FAS 35.6±8 words vs. 40±13 words, respectively; $p>0.3$ for all measures).

Single-task standing: total spectral power

During the usual single-task standing condition (eyes open/hard surface), there was no difference between the two groups for total spectral power in the ML ($p=0.3$) or AP ($p=0.1$) direction. In the ML direction, greater postural challenge significantly increased spectral power in both groups (Vision x Surface interaction $p<0.001$; Figure 1a, solid lines). Similarly, in the AP direction, greater postural challenge increased spectral power in both groups (Vision x Surface interaction p=0.005; Figure 1b, solid lines), however prosthesis users had a greater increase in spectral power (Vision x Group interaction p=0.020; Surface x Group interaction p=0.036). These results support the hypothesis that the greatest difference between groups would occur in the AP direction (hypothesis 1).

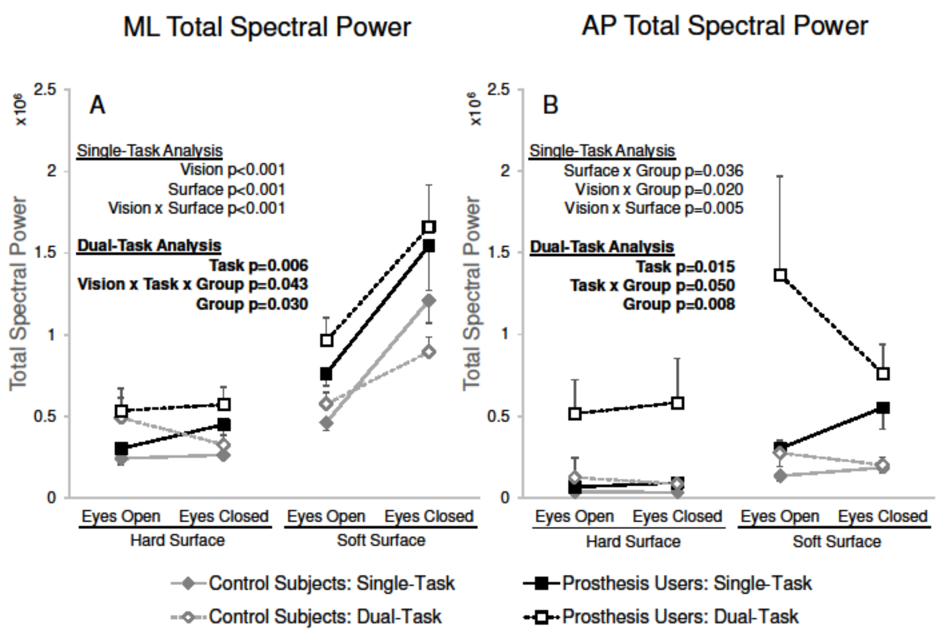

Fig. 1. Single-task and Dual-task total spectral power (mean and standard error) in the ML (a) and AP (b) directions with increasing postural difficulty (eyes closed and soft surface). During single-task (solid lines), prosthesis users (black squares) and control subjects (gray diamonds) increased spectral power with increasing postural challenge in both the AP and ML directions; prosthesis users had a greater increase than control subjects in the AP direction. Dual-tasking (dashed lines) had a differential effect between groups with prosthesis users having a greater increase than control subjects.

Single-task standing: frequency band contributions

Tables 1 and 2 show the relative spectral power from each frequency band during the single-task standing conditions and associated statistics. In both the ML and AP directions, the eyes closed conditions significantly reduced the relative spectral power from the very low frequency band (main effect Vision: ML $p=0.005$; AP $p<0.001$). Further, during the eyes closed conditions there was an increase in the relative spectral power from the middle (main effect Vision: ML $p<0.001$; AP $p=0.001$) and high (main effect Vision: ML $p<0.001$; AP $p<0.001$) frequency bands. Standing on the soft surface also significantly increased the relative spectral power from the middle frequency band in the ML direction (main effect Surface $p=0.001$) There was a similar trend in the AP direction (main effect Surface $p=0.030$). The soft surface also increased the contribution from the low frequency band (main effect Surface: ML $p=0.003$; AP $p=0.001$) while reducing the very low frequency band contribution (main effect Surface: ML $p<0.001$; AP $p=0.003$). These results support hypotheses 2A (a decrease in the relative contribution from the very low frequency band with eyes closed) and 2B (an increase in the relative contribution from the middle frequency band on soft surface). There were no significant differences between groups for relative spectral power (main effect Group and interactions: ML $p=0.495$-0.970; AP $p=0.032$-0.961, appendix Tables I and II).

Dual-task standing: total spectral power

The cognitive task had a differential effect on total spectral power between the groups. In the ML direction, there was a Vision x Task x Group interaction ($p=0.043$; Figure 1a) indicating that while dual-tasking with eyes closed prosthesis users increased the total spectral power while control subjects did not. In the AP direction, there was a Task x Group interaction ($p=0.050$; Figure 1b) showing that prosthesis users had a significantly higher total spectral power while dual-tasking compared to control subjects. These results support a differential response to dual-tasking between groups for the total spectral power (hypothesis 3).

Table 1
The relative spectral power (%) from different frequency bands (mean and SE) in the ML direction during single-task and dual-task standing. The standing condition and the cognitive task both impacted the relative contribution from each frequency band. There was no difference between groups in relative contribution. Interactions were not significant (p>0.03).

Standing Condition	Task	Very Low [0 – 0.19 Hz)		Low [0.19 – 0.39 Hz)		Middle [0.39 – 1.17 Hz)		High [1.17 – 10.15 Hz)	
		Prosthesis	Control	Prosthesis	Control	Prosthesis	Control	Prosthesis	Control
Eyes Open/Hard Surface									
	Single	75 (3)	75 (3)	12 (2)	12 (2)	8 (1)	9 (1)	4 (1)	4 (1)
	Dual	59 (7)	59 (6)	12 (3)	17 (4)	17 (3)	16 (3)	11 (3)	8 (2)
Eyes Closed/Hard Surface									
	Single	70 (3)	68 (5)	11 (1)	10 (1)	11 (1)	13 (2)	7 (1)	8 (2)
	Dual	63 (4)	59 (6)	9 (2)	15 (3)	17 (2)	16 (3)	11 (2)	9 (2)
Eyes Open/Soft Surface									
	Single	67 (4)	67 (3)	18 (3)	16 (2)	11 (1)	11 (1)	4 (1)	5 (1)
	Dual	60 (5)	56 (4)	24 (3)	24 (2)	11 (2)	14 (3)	5 (1)	5 (1)
Eyes Closed/Soft Surface									
	Single	58 (4)	57 (5)	16 (2)	16 (2)	17 (2)	17 (2)	8 (1)	10 (3)
	Dual	49 (5)	52 (6)	19 (2)	20 (3)	20 (3)	19 (3)	12 (1)	9 (2)
ANOVA Single-Task									
Main Effects	Vision	**0.005**		0.417		**<0.001**		**<0.001**	
	Surface	**<0.001**		**0.003**		**0.001**		0.080	
	Group	0.876		0.614		0.702		0.580	
ANOVA Dual-Task									
Main Effect	Task	**<0.001**		**0.004**		**0.001**		**0.009**	

Table 2
The relative spectral power (%) from different frequency bands (mean and SE) in the AP direction during single-task and dual-task standing. The standing condition impacted the relative contribution from each frequency band, however, performing the cognitive task did not. There was no difference between groups in relative contribution. Interactions were not significant (p>0.05).

Standing Condition	Task	Very Low [0 – 0.19 Hz)		Low [0.19 – 0.39 Hz)		Middle [0.39 – 1.17 Hz)		High [1.17 – 10.15 Hz)	
		Prosthesis	Control	Prosthesis	Control	Prosthesis	Control	Prosthesis	Control
Eyes Open/Hard Surface									
	Single	70 (3)	76 (3)	9 (1)	8 (2)	14 (2)	11 (1)	6 (1)	5 (1)
	Dual	67 (7)	75 (5)	10 (2)	12 (4)	17 (4)	9 (2)	6 (2)	5 (1)
Eyes Closed/Hard Surface									
	Single	62 (5)	70 (4)	8 (2)	9 (2)	18 (3)	13 (2)	12 (3)	8 (1)
	Dual	65 (6)	69 (6)	10 (2)	7 (2)	15 (4)	13 (3)	8 (2)	10 (4)
Eyes Open/Soft Surface									
	Single	63 (5)	66 (5)	13 (3)	17 (3)	17 (3)	13 (3)	7 (2)	3 (1)
	Dual	59 (7)	65 (5)	18 (3)	17 (2)	16 (3)	13 (3)	6 (2)	3 (1)
Eyes Closed/Soft Surface									
	Single	46 (5)	56 (6)	18 (3)	17 (2)	24 (2)	20 (4)	12 (2)	7 (1)
	Dual	62 (5)	60 (6)	14 (3)	17 (3)	15 (3)	18 (4)	8 (2)	5 (1)
ANOVA Single-Task									
Main Effects	Vision	**<0.001**		0.278		**0.001**		**<0.001**	
	Surface	**0.003**		**0.001**		**0.030**		0.813	
	Group	0.107		0.780		0.091		0.032	
ANOVA Dual-Task									
Main Effect	Task	0.597		0.491		0.307		0.410	

Dual-task standing: frequency band contributions

Tables 1 and 2 show the relative spectral power from each frequency band while dual-tasking and associated statistics. Performing a cognitive task did not have a significant effect on relative contributions between selected frequency bands in the AP direction (main effect Task and interactions $p \geq 0.056$, Table 2 and Appendix B Table II). In the ML direction there were no differences between groups (main effect Group and interactions $p > 0.086$, Table 1 and Appendix B Table I), however, the cognitive task did impact the relative contribution between frequency bands. Performing a cognitive task significantly reduced the contribution from the very low frequency band across standing conditions (main effect Task: $p < 0.001$). There was also an overall increase in the low, middle and high frequency bands (main effect Task $p = 0.001$-0.009). Thus, although dual-tasking does impact the relative contribution from each band in the ML direction, the lack of a differential response between groups in either direction fails to support hypothesis 3 for relative spectral power.

Bilateral comparisons: total spectral power

Prosthesis users exhibited an asymmetry in total spectral power between the prosthetic and intact sides in both the ML (Figure 2) and AP (Figure 3) directions with more power on the intact side. In the ML direction, asymmetry increased with the more challenging standing conditions (Figure 2, main effect Side $p < 0.001$; Surface x Side interaction $p < 0.001$; Vision x Side interaction $p = 0.017$; Vision x Surface x Side interaction $p = 0.002$). However, in the AP direction significant asymmetry was only present while standing on the soft surface (Figure 3, main effect Side $p = 0.084$; Surface x Side interactions $p = 0.044$).

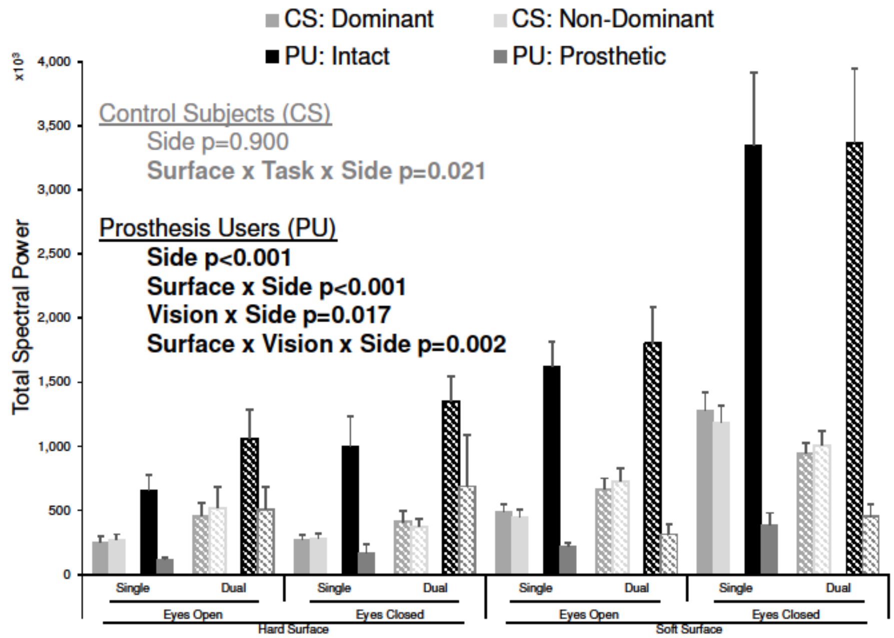

Fig. 2. Total spectral power in the ML direction (mean and standard error) for each side for prosthesis users (PU) and control subjects (CS) during single-task (solid) and dual-task (hatched) standing. The prosthesis users had more power on the intact side (black) than the prosthetic side (dark gray) across standing conditions. However, the asymmetry increased in the more challenging standing conditions. The control subjects did not exhibit asymmetry between the dominant (gray) and non-dominant (light gray) sides.

Control subjects did not exhibit significant asymmetry between the dominant and non-dominant side in the ML direction (Figure 2, main effect Side p=0.900). However, during the soft surface standing condition there was a small but significant switch from the dominant to non-dominant side having more power while dual-tasking (Figure 2, Surface x Task x Side interaction p=0.021). The dominant side did have higher spectral power in the AP direction (Figure 3, main effect Side: p=0.005) with the greatest difference in the most difficult, single task standing condition (Figure 3, Vision x Surface x Task x Side interaction p=0.028). The asymmetry in the eyes open only or hard surface only condition was confirmed with a follow up analysis evaluating the effect of Side without the eyes closed or soft surface condition (2x2x2 ANOVA Side p=0.009 and p=0.024, respectively).

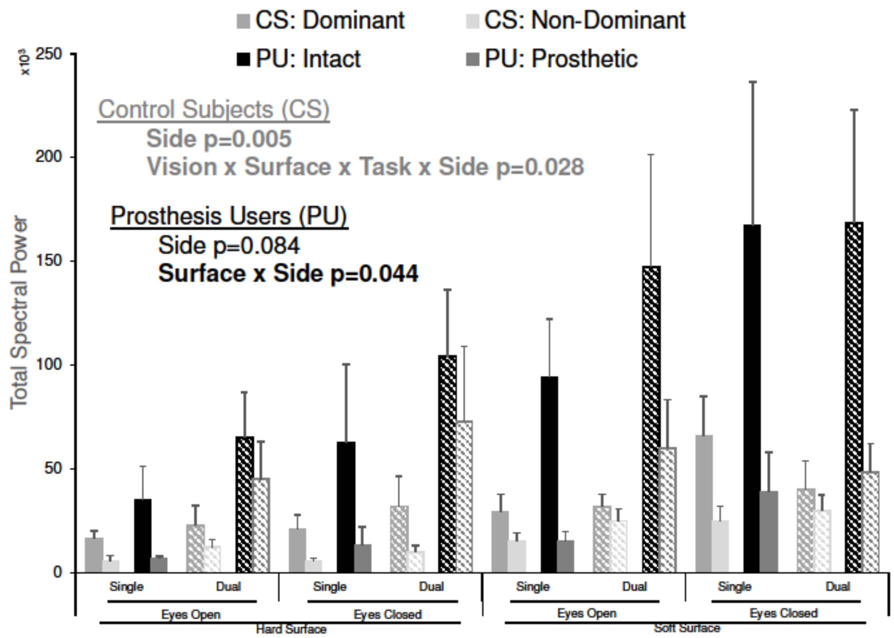

Fig. 3. Total spectral power in the AP direction (mean and standard error) for each side for prosthesis users (PU) and control subjects (CS) during single-task (solid) and dual-task (hatched) standing. The prosthesis users only exhibited a significant asymmetry during the soft surface standing condition with more power on the intact side (black) than the prosthetic side (dark gray). The control subjects had more power on the dominant (gray) side than the non-dominant (light gray) side across standing conditions. The largest differences between sides was during the eyes closed/soft surface, single-task standing condition.

Bilateral comparisons: frequency band relative spectral power

There was a difference between the prosthetic and intact sides in the relative contribution from the very low, middle, and high frequency bands in the ML direction (Table 3). The intact side had a smaller contribution from the very low frequency band than the prosthetic side (main effect Side p<0.001). Conversely, the prosthetic side had a smaller contribution from the middle (Figure 4) and high frequency bands than the intact side (main effect Side p<0.001, both). Overall, dual-tasking did not impact the difference in relative spectral power between sides (Task x Side interactions p≥0.068, appendix Table III) except for the high frequency band in the ML direction (Vision x Task x Side interaction p=0.012, Appendix B Table III). In the AP direction only the high frequency band showed a difference between the prosthetic and intact side, with greater relative contribution on the intact side (main effect Side p=0.022, Appendix B Table IV).

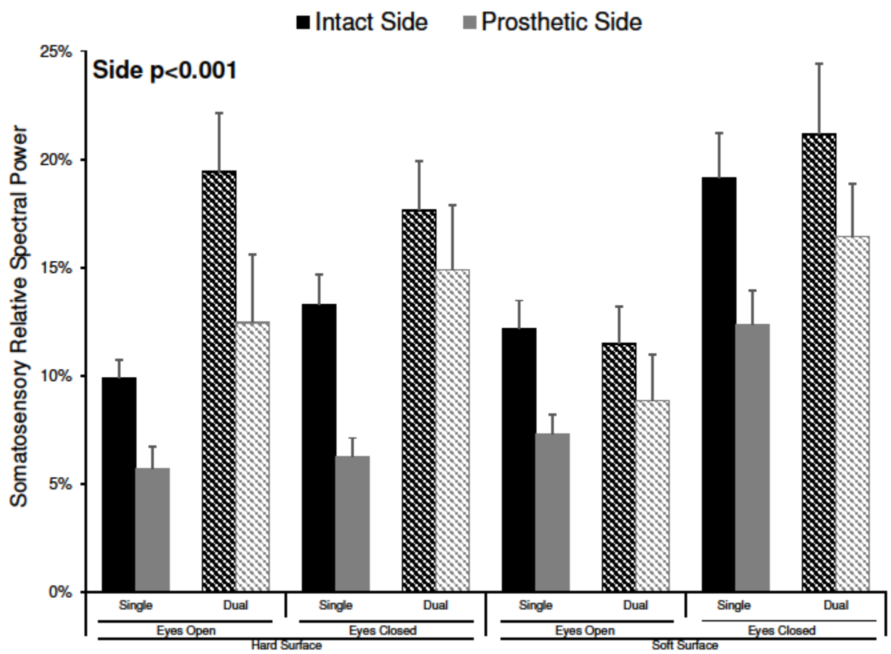

Fig. 4. Relative spectral power from the middle frequency band (mean and standard error) in the ML direction for the intact (black) and prosthetic (gray) sides during single-task (solid) and dual-task (hatched) standing. The prosthetic side had a smaller contribution from the middle frequency band across all standing conditions.

Table 3

The relative spectral power (%) from different frequency bands (mean and SE) recorded on the prosthetic and intact sides in the ML direction during single-task and dual-task standing. The prosthetic side had lower contribution from the middle and high frequency bands and higher contribution from the very low frequency band. Only the high frequency band had a significant Side interactions (Vision x Side p=0.010; Vision x Task x Side p=0.012); showing less asymmetry while dual-tasking with eyes closed. Remaining Side interactions were not significant (p>0.08).

Standing Condition	Task	Very Low [0 – 0.19 Hz)		Low [0.19 – 0.39 Hz)		Middle [0.39 – 1.17 Hz)		High [1.17 – 10.15 Hz)	
		Prosthetic	Intact	Prosthetic	Intact	Prosthetic	Intact	Prosthetic	Intact
Eyes Open/Hard Surface									
	Single	81 (3)	72 (3)	12 (2)	13 (2)	6 (1)	10 (1)	1 (0.2)	5 (1)
	Dual	68 (7)	53 (7)	11 (3)	14 (3)	12 (3)	19 (3)	9 (3)	14 (3)
Eyes Closed/Hard Surface									
	Single	79 (2)	67 (3)	12 (2)	10 (1)	6 (1)	13 (1)	3 (0.4)	9 (1)
	Dual	66 (5)	62 (5)	9 (2)	9 (2)	15 (3)	18 (2)	10 (3)	11 (2)
Eyes Open/Soft Surface									
	Single	73 (4)	64 (4)	17 (3)	18 (3)	7 (1)	12 (1)	2 (0.3)	6 (1)
	Dual	69 (5)	58 (5)	19 (4)	24 (4)	9 (2)	11 (2)	4 (1)	6 (1)
Eyes Closed/Soft Surface									
	Single	67 (4)	55 (4)	16 (2)	16 (2)	12 (2)	19 (2)	4 (1)	10 (1)
	Dual	58 (5)	48 (6)	18 (3)	18 (3)	16 (2)	21 (3)	7 (1)	13 (1)
ANOVA									
Main Effect	Side	**<0.001**		0.228		**<0.001**		**<0.001**	

Discussion

This study provides new insight into the use of sensory systems for balance maintenance in prosthesis users. The results support the findings of our previous analysis of temporal-spatial sway parameters in each group [3] while also characterizing the spectral features of sway when sensory systems are limited or stressed. Prosthesis users had a greater increase in spectral power than control subjects in the AP direction during the more difficult standing conditions. Altering the sensorimotor demands of the standing conditions changed the spectral characteristics similarly in both groups. The frequency band associated with vision [7, 9] had a smaller contribution with eyes closed. Standing on a soft surface increased the contributions from the somatosensory frequency band [4, 7]. Dual-tasking resulted in a greater increase in spectral power for prosthesis users but not control subjects in both the ML and AP directions. Dual-tasking also lowered the contribution from the band associated with vision (very low frequency band) while increasing the contributions from the other bands in the ML direction in both groups. Prosthesis users exhibited asymmetry in spectral power with more power on the intact side in both the ML and AP directions. There was also a smaller contribution from the somatosensory frequency band on the prosthetic side. These results suggest that prosthesis users have greater limitations in postural control in the AP than ML direction and are overall more disrupted while dual-tasking. However, dual-tasking impacts the use of sensory systems in postural control for both prosthesis users and control subjects similarly. Further, the overall sensory contributions to postural control are similar between groups but dominated by the intact side in prosthesis users with less somatosensory control on the prosthetic side.

The total spectral power results closely mirror the temporal-spatial sway characteristics reported in [3]. Specifically, no differences were observed between groups in the eyes open/hard surface standing condition and there was greater increase in prosthesis users in total spectral power during the more difficult standing conditions, either physically imposed or due to the cognitive task. These results support findings that more challenging standing conditions may be required to observe differences in postural control between prosthesis users and control subjects [26, 27]. The total spectral power results also support the hypothesis that postural control in the AP direction is more disrupted than the ML, presumably due to the limited function of the prosthetic ankle. Further, the differential dual-task results between prosthesis users and control subjects support the finding that postural control in prosthesis users is more strongly affected by increased cognitive demands. This behavior is likely due to increased cognitive demand imposed by use of a prosthesis, which is supported by subjective reports [28, 29], and the apparent prioritization to achieve desired goals despite increased risk [3].

Wavelet analysis was able to characterize the spectral characteristics in the selected frequency bands and the changes in the relative spectral power from the frequency bands were consistent with the nature of the challenge in the various standing conditions. The contribution from the very low frequency band [0 – 0.19 Hz), which is associated with vision [7, 9], was reduced in the eyes closed condition with a subsequent increase in the other frequency bands (hypothesis 2A). Further, the soft surface standing conditions resulted in an increase in the contribution from the middle frequency band [0.39 – 1.17 Hz), which is associated with somatosensory control (hypothesis 2B) [4, 7]. Confirmation of the predicted results support other studies, which found that specific frequency bands in the CoP signal are associated with specific sensory systems [4, 7, 9]. Further, these results support a complex model of postural control that relies on the coordination of multiple sensory systems [30, 31] as well as versatility in recruiting these sensory systems to respond to specific postural control demands [32].

The interpretation of the results relies on the theory that each sensory system is primarily associated with one frequency band. However, this model of postural control does not require that one sensory system exclusively drives postural adjustments captured in the specified spectral range. For example, as in other studies [7, 9, 13], the power in the very low frequency band did not totally disappear in the eyes closed condition. While visual control is primarily associated with the lowest frequencies, other factors or systems also contribute to this band. This is likely the case with each frequency band and certainly the spectral borders suggested for each system serve as guidelines rather than precise boundaries. Most studies use group means to establish the sensory associated frequency bands; however, when individual results are reported there is evidence of between-subject differences in spectral boundaries [14, 16]. Nevertheless, the consistency in results across several studies still support use of these general guidelines.

Dual-tasking impacted the spectral characteristics of the CoP signal in the ML direction only and reduced the relative contribution of the frequency band primarily attributed to vision across all standing conditions. Kirchner et al. and Chagdes et al. also found dual-tasking to reduce the contribution from the vision frequency band for visual/memory and motor tasks, respectively [8, 9]. While both sets of authors attribute the decrease contribution from this frequency band to sensory reweighting, the justification is based primarily on the nature of the task. For example, in Kirchner et al., the additional task required subjects to look at images projected on a wall, and the authors suggested that the reduction in the lowest frequencies was due to vision being engaged elsewhere [8]. However, the consistent finding of reduced very low frequency contribution across notably different dual-tasks suggests a more general, rather than task-specific, mechanism for sensory reweighting. The visual system is highly engaged in conscious activities and therefore may be most susceptible to competing demands from other conscious activities. This fits well with the central capacity-sharing theory [33, 34] and the similar cross-domain completion model [2] which propose that simultaneous performance of tasks competing for the same resources would decrease performance in one or both tasks. In contrast, the level of alertness hypothesis, which suggests increased cognitive resource recruitment with increased demand [35, 36], may be supported by the increased contribution of the somatosensory system (middle frequency band), while standing on the soft surface. When the somatosensory system was stressed, rather than removed, more resources were allocated to that system. Perhaps both models work in conjunction to explain dual-task behavior. While more resources, or attention, may be applied to a system in response to higher demand, there remains a limitation to capacity. Dual-tasking may increase the competition for resources in some systems more than others. Increasing the allocation of resources from the sensory system in high demand or from other supporting systems may be sufficient to meet the postural challenge. Thus, central capacity-sharing may only be evident when contributions from specific sensory systems are evaluated.

An evaluation of the spectral characteristics between the prosthetic and intact sides revealed asymmetries in both the total spectral power and the relative spectral power from some frequency bands. The intact side had higher total spectral power than the prosthetic side, which was more notable in the ML direction. Control subjects also had an asymmetric total power distribution, however it was only in the AP direction and the asymmetry was less pronounced. These results fit well with other reports of asymmetry in prosthesis users that found greater reliance on the intact side [37, 38]. Within the frequency bands, the prosthetic side had a smaller contribution from the somatosensory band. Considering that the somatosensory system is directly impacted by amputation, the lower contribution from the middle frequency band further supports reports of somatosensory association with that band [4, 7]. As the somatosensory frequency band contribution was not different from control subjects in the resultant CoP signal analysis, it also appears that the intact side is capable of compensating for the impaired system. These results may allow for quantification of the impact of different prosthetic devices on somatosensory control of posture. Specifically, similar methods could be used to assess how a dynamically controlled prosthesis or one that provides sensory feedback specifically impacts the use of the somatosensory system.

Limitations

One limitation of wavelet analysis is the lack of specific guidelines on selecting the most appropriate mother wavelet [25]. However, Daubechies 4 met all necessary requirements and there were no major differences in the results produced by additional analysis with a different candidate mother wavelet (Coiflets). In this analysis, the resolution of the frequency bands, particularly the lowest frequency bands, could have been improved by longer trial durations [8]. Longer trials would have also allowed the frequency band resolution to better align with the suggested boundaries between regions for the sensory systems. In addition, a larger sample size may provide more robust evidence of the spectral characteristics and a more homogeneous sample with respect to age, the cause of amputation, or type of componentry may have provided more insight into specific drivers of changes in postural control. However, the suitability of the subject pool is supported by the fact these results followed predicted outcomes based upon studies in other populations.

Conclusion

Wavelet analysis successfully characterized the spectral features of the CoP signal in prosthesis users and control subjects during single- and dual-task standing. The groups had similar changes in the frequency composition of the total spectral power during the different standing conditions. The changes in response to the eyes closed and soft surface conditions fit well with the suggested frequency bands associated with the visual and somatosensory systems suggesting that the contributions from these systems can be characterized. The results also show that the contribution from the somatosensory system is reduced on the prosthetic side. This observation may form the basis for evaluating the effectiveness of prosthetic devices aimed at improving sensory feedback and control. Dual-tasking reduced the contribution from the very low frequency band, and as this is consistent with other studies; this finding may provide insight into the mechanisms that individuals use to navigate the performance of multiple tasks. Analysis of total spectral power supports previous conclusions that postural control in the AP direction is more disrupted in prosthesis users than the ML with increasing postural challenge and that prosthesis users are more impacted by performing a cognitive task than control subjects [3]. Devices and strategies that aim to reduce the cognitive burden of using a prosthesis and provide somatosensory feedback may improve postural control in below-knee prosthesis users.

References

[1] Woollacott M, Shumway-Cook A. Attention and the control of posture and gait: a review of an emerging area of research. Gait Posture. 2002; 16:1-14.

[2] Lacour M, Bernard-Demanze L, Dumitrescu M. Posture control, aging, and attention resources: Models and posture-analysis methods. Clin Neurophysiol. 2008; 38:411-21.

[3] Howard CL, Perry B, Chow JW, Wallace C, Stokic DS. Increased alertness, better than posture prioritization, explains dual-task performance in prosthesis users and controls under increasing postural and cognitive challenge. Exp Brain Res. 2017; Epub: Aug 31; 10.1007/s00221-017-5077-2.

[4] Oppenheim U, Kohen-Raz R, Alex D, Kohen-Raz A, Azarya M. Postural characteristics of diabetic neuropathy. Diabetes care. 1999; 22:328-32.

[5] Mauritz KH, Dichgans J, Hufschmidt A. Quantitative analysis of stance in late cortical cerebellar atrophy of the anterior lobe and other forms of cerebellar ataxia. Brain. 1979; 102:461-82.

[6] Diener HC, Dichgans J, Bacher M, Gompf B. Quantification of postural sway in normals and patients with cerebellar diseases. Electroencephalogr Clin Neurophysiol. 1984; 57:134-42.

[7] Thurner S, Mittermaier C, Hanel R, Ehrenberger K. Scaling-violation phenomena and fractality in the human posture control systems. Phys Rev E Stat Phys Plasmas Fluids Relat Interdiscip Topics. 2000; 62:4018-24.

[8] Kirchner M, Schubert P, Schmidtbleicher D, Haas C. Evaluation of the temporal structure of postural sway fluctuations based on a comprehensive set of analysis tools. Phys A. 2012; 391:4692-703.

[9] Chagdes JR, Rietdyk S, Haddad JM, Zelaznik HN, Raman A, Rhea CK, et al. Multiple timescales in postural dynamics associated with vision and a secondary task are revealed by wavelet analysis. Exp Brain Res. 2009; 197:297-310.

[10] Bernard-Demanze L, Dumitrescu M, Jimeno P, Borel L, Lacour M. Age-related changes in posture control are differentially affected by postural and cognitive task complexity. Curr Aging Sci. 2009; 2:139-49.

[11] Mauritz KH, Dietz V. Characteristics of postural instability induced by ischemic blocking of leg afferents. Exp Brain Res. 1980; 38:117-9.

[12] Diener HC, Dichgans J, Bruzek W, Selinka H. Stabilization of human posture during induced oscillations of the body. Exp Brain Res. 1982; 45:126-32.

[13] Diener HC, Dichgans J, Guschlbauer B, Mau H. The significance of proprioception on postural stabilization as assessed by ischemia. Brain Res. 1984; 296:103-9.

[14] Collins JJ, De Luca CJ. Open-loop and closed-loop control of posture: a random-walk analysis of center-of-pressure trajectories. Exp Brain Res. 1993; 95:308-18.

[15] Collins JJ, De Luca CJ. Random walking during quiet standing. Phys Rev Lett. 1994; 73:764-7.

[16] Singh NK, Snoussi H, Hewson D, Duchêne J. Wavelet transform analysis of the power spectrum of centre of pressure signals to detect the critical point interval of postural control. Biomedical Engineering Systems and Technologies: Springer; 2010. p. 235-44.

[17] Roeing KL, Wajda DA, Sosnoff JJ. Time dependent structure of postural sway in individuals with multiple sclerosis. Gait Posture. 2016; 48:19-23.

[18] Pajala S, Era P, Koskenvuo M, Kaprio J, Tormakangas T, Rantanen T. Force platform balance measures as predictors of indoor and outdoor falls in community-dwelling women aged 63-76 years. J Gerontol A Biol Sci Med Sci. 2008; 63:171-8.

[19] Maatar D, Fournier R, Lachiri Z, Nait-Ali A. Discrete wavelet and modified PCA decompositions for postural stability analysis in biometric applications. J Biomed Sci Eng. 2011; 4:543.

[20] Sample RB, Jackson K, Kinney AL, Diestelkamp WS, Reinert SS, Bigelow KE. Manual and Cognitive Dual Tasks Contribute to Fall-Risk Differentiation in Posturography Measures. J Appl Biomech. 2016; 32:541-7.

[21] Martinez-Ramirez A, Lecumberri P, Gomez M, Izquierdo M. Wavelet analysis based on time-frequency information discriminate chronic ankle instability. Clin Biomech. 2010; 25:256-64.

[22] Martinez-Ramirez A, Lecumberri P, Gomez M, Rodriguez-Manas L, Garcia FJ, Izquierdo M. Frailty assessment based on wavelet analysis during quiet standing balance test. J Biomech. 2011; 44:2213-20.

[23] In F, Kim S. Introduction to Wavelet Theory in Finance : A Wavelet Multiscale Approach. Singapore, UNITED STATES: World Scientific Publishing Company; 2012.

[24] Howard CL, Wallace C, Abbas J, Stokic DS. Residual standard deviation: Validation of a new measure of dual-task cost in below-knee prosthesis users. Gait Posture. 2017; 51:91-6.

[25] Torrence C, Compo GP. A practical guide to wavelet analysis. Bull Amer Meteor. 1998; 79:61-78.

[26] Vanicek N, Strike S, McNaughton L, Polman R. Postural responses to dynamic perturbations in amputee fallers versus nonfallers: a comparative study with able-bodied subjects. Arch Phys Med Rehabil. 2009; 90:1018-25.

[27] Mohieldin A, Chidambaram A, Sabapathivinayagam R, Al Busairi W. Quantitative assessment of postural stability and balance between persons with lower limb amputation and normal subjects by using dynamic posturography. Maced J Med Sci. 2010; 3:138-43.

[28] Vrieling AH, van Keeken HG, Schoppen T, Otten E, Hof AL, Halbertsma JP, et al. Balance control on a moving platform in unilateral lower limb amputees. Gait Posture. 2008; 28:222-8.

[29] Morgan SJ, Hafner BJ, Kelly VE. The effects of a concurrent task on walking in persons with transfemoral amputation compared to persons without limb loss. Prosthet Orthot Int. 2016; 40:490-6.

[30] Morasso PG, Schieppati M. Can muscle stiffness alone stabilize upright standing? J Neurophysiol. 1999; 82:1622-6.

[31] Morasso PG, Sanguineti V. Ankle muscle stiffness alone cannot stabilize balance during quiet standing. J Neurophysiol. 2002; 88:2157-62.

[32] Peterka R. Sensorimotor integration in human postural control. Journal of neurophysiology. 2002; 88:1097-118.

[33] Beurskens R, Steinberg F, Antoniewicz F, Wolff W, Granacher U. Neural correlates of dual-task walking: Effects of cognitive versus motor interference in young adults. Neural Plast. 2016; 2016:1-9.

[34] Tombu M, Jolicoeur P. A central capacity sharing model of dual-task performance. J Exp Psychol Hum Percept Perform. 2003; 29:3-18.

[35] Bonnet CT, Baudry S. Active vision task and postural control in healthy, young adults: Synergy and probably not duality. Gait Posture. 2016; 48:57-63.

[36] Bisson EJ, Lajoie Y, Bilodeau M. The influence of age and surface compliance on changes in postural control and attention due to ankle neuromuscular fatigue. Exp Brain Res. 2014; 232:837-45.

[37] Highsmith MJ, Schulz BW, Hart-Hughes S, Latlief GA, Phillips SL. Differences in the spatiotemporal parameters of transtibial and transfemoral amputee gait. J Prosthet Orthot. 2010; 22:26-30

[38] Ku PX, Abu Osman NA, Wan Abas WA. Balance control in lower extremity amputees during quiet standing: A systematic review. Gait Posture. 2014; 39:672-82.

CHAPTER 7

RESIDUAL STANDARD DEVIATION: VALIDATION OF A NEW MEASURE OF DUAL-TASK COST IN BELOW-KNEE PROSTHESIS USERS

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Howard C, Wallace C, Abbas J, Stokic DS. Residual standard deviation: Validation of a new measure of dual-task cost in below-knee prosthesis users. Gait Posture. 2017; 51: 91-96. 10.1016/j.gaitpost.2016.09.025
http://www.sciencedirect.com/science/article/pii/S0966636216305872

An unpublished derivation of the mathematical theory of the presented method is provided in Appendix C.

Abstract

We developed and evaluated properties of a new measure of variability in stride length and cadence, termed residual standard deviation (RSD). To calculate RSD, stride length and cadence are regressed against velocity to derive the best fit line from which the variability (SD) of the distance between the actual and predicted data points is calculated. We examined construct, concurrent, and discriminative validity of RSD using dual-task paradigm in 14 below-knee prosthesis users and 13 age- and education-matched controls. Subjects walked first over an electronic walkway while performing separately a serial subtraction and backwards spelling task, and then at self-selected slow, normal, and fast speeds used to derive the best fit line for stride length and cadence against velocity. Construct validity was demonstrated by significantly greater increase in RSD during dual-task gait in prosthesis users than controls (group-by-condition interaction, stride length $p=0.0006$, cadence $p=0.009$). Concurrent validity was established against

coefficient of variation (CV) by moderate-to-high correlations (r=0.50-0.87) between dual-task cost RSD and dual-task cost CV for both stride length and cadence in prosthesis users and controls. Discriminative validity was documented by the ability of dual-task cost calculated from RSD to effectively differentiate prosthesis users from controls (area under the receiver operating characteristic curve, stride length 0.863, p=0.001, cadence 0.808, p=0.007), which was better than the ability of dual-task cost CV (0.692, 0.648, respectively, not significant). These results validate RSD as a new measure of variability in below-knee prosthesis users. Future studies should include larger cohorts and other populations to ascertain its generalizability.

Introduction

Effective control of gait requires complex coordination of multiple joints, limb segments, and muscles through various sensory-motor mechanisms. These control mechanisms modulate propulsion, braking, and body support during a gait cycle in response to ambulation goals and environmental demands. Despite the many complex mechanisms engaged, the resultant gait characteristics form consistent patterns of coordination. Most notably, stride length and cadence are modulated together forming a strong linear relationship along a broad range of gait speeds [1, 2]. However, environmental influences and limitations inherent to human sensory-motor control introduce variability, which is apparent in healthy subjects and exaggerated after a neurological or musculoskeletal injury. For example, the strength of the linear relationship between stride length and cadence is weaker in Parkinson’s disease [3] and prosthesis users [4] compared to unimpaired controls.

Disturbed sensory-motor control of gait in prosthesis users may be ascribed to a loss of limb, impaired sensation, or current limitations of prosthetic devices. This requires engaging additional motor and cognitive resources that impose load during performance of daily tasks. Not surprisingly, therefore, prosthesis users prefer componentry that they perceive less cognitively demanding [5-7]. The demand is amplified by frequent presence of cognitive impairments in prosthesis users [8].

Cognitive-motor interference is commonly induced with a dual-task paradigm, which requires performance of an additional task while walking. The increased load on the sensory-motor system typically alters gait and has been related to fall risk and instability [9-11]. Dual-task gait has more ecological validity than typical gait analysis and may elicit deviations not seen during regular walking [12, 13]. Despite greater ecological validity and potential for improving sensitivity of gait studies, dual-task gait has not been extensively studied in prosthesis users. Some studies only looked at the cognitive performance [5, 6], whereas others reported no significant increase in cognitive-motor interference in above-knee prosthesis users [14, 15]. The reported absence of interference in the above-knee prosthesis users may be due to a small sample size, concurrent task selection, instructions about prioritization, substantial gait deviations in the single-task condition that constrained emergence of further perturbation under dual-tasking in order to preserve stability, or insensitive outcome measures.

The most commonly reported outcome in dual-task gait studies is the variability of temporal-spatial parameters [16, 17]. The selected index of variability, however, is not uniformly defined or clearly justified with respect to studied gait parameters or experimental designs. Some studies use the standard deviation (SD) because it requires little data manipulation, thus, simplifying interpretation [17-19]. Most investigators report the coefficient of variation (CV), the ratio (%) of SD to the mean value of the parameter of interest. When examining variability, the relationship of gait parameters with velocity is typically not considered. However, because such relationships commonly exist, the parameter mean and SD are not independent of, or proportionally scaled with, velocity [18-20]. Thus, spontaneous or induced fluctuations in velocity may variably affect SD and mean values, leading to ambiguity in interpretation. This especially pertains to dual-task studies, because addition of a concurrent task tends to decrease velocity and alter related gait parameters [12, 15, 21]. Thus, there is a need to account for the impact of velocity on gait parameters for which the measures of variability are derived.

To control for velocity between conditions, previous studies have used a treadmill [22], analyzed data that fell within a narrow range of the prescribed speed [23], or made mathematical adjustments [24, 25]. For example, Nordin et al.[25] used the linear relationship that step length and step time have with velocity to predict their values across a range of speeds and calculate the difference between the actual mean values and the predicted values for each condition. This reportedly improved detection of a dual-task cost (difference between single- and dual-task conditions). Because the mean values were used for calculating the dual-task cost, it was not possible to derive variability across multiple gait cycles. To overcome this, we extend the above approach by proposing a new method for analyzing variability in stride length and cadence that takes into account their close relationship with velocity. We termed this new index of variability the residual standard deviation (RSD), because it calculates a SD of the *vertical distance* between each actual data point and the point predicted by the best fit line between the velocity and stride length/cadence. Thus, RSD quantifies the variability of a departure from the linear relationship that stride length and cadence have with velocity across the range of self-selected walking speeds.

The purpose of this study was to validate the RSD method for calculating variability of stride length and cadence. For construct validity (aim 1), we compared changes in RSD from baseline to dual-task gait between below-knee prosthesis users and age/education-matched non-amputee controls. Aside from rare dual-task studies in this population, this choice was guided by our recent findings of the disrupted stride length-cadence relationship in below-knee prosthesis users [4]. The reduced automaticity (i.e., more variable sensory-motor output) was expected to be exaggerated during dual-task gait and captured by RSD. Concurrent validity (aim 2) was examined by correlating dual-task cost RSD with dual-task cost CV to infer to which degree the two measures probe the same construct. Discriminant validity (aim 3) was evaluated by examining the ability of dual-task cost RSD to differentiate below-knee prosthesis users from controls. As a follow-up, the discriminant ability of dual-task cost RSD was compared to the same ability of dual-task cost CV. Our first hypothesis was that RSD will capture larger variability in both stride length and cadence during dual-task gait in below-knee prosthesis users compared to controls. The second hypothesis was that dual-task cost RSD will positively and at least moderately correlate with dual-task cost CV. The third hypothesis was that the receiver operating characteristics (ROC) analysis based on dual-task cost RSD will yield a significant area under the curve (AUC) when comparing prosthesis users to controls.

Methods

Participants

A convenience sample of unilateral below-knee prosthesis users was recruited from clinics run by our institution. The inclusion criteria were ≥ 1 year since amputation; age 18–80 years; comfortable socket fit; no known balance, neurological, or other health problems that limit daily activities; and able to safely walk 10m-distance at different velocities, as verified by a certified prosthetist. Age- and education-matched non-amputee subjects were recruited from the community to serve as controls. While not specifically matched for gender, we recruited more male subjects in the control sample to better approximate the prosthesis user population [26].

The sample included 13 controls (mean age 46±18 years, 15±2 years of education, BMI 26±3, 8 men) and 14 below-knee prosthesis users (age 43±12 years, 14±2 years of education, BMI 26±3, 11 men). The difference in the proportions of male vs. female subjects in the two samples was not significant (Fischer exact test, p=0.420). The amputation occurred 9±7 years (1.0 to 28) earlier due to trauma (n=11), infection (n=2), or vascular disease (n=1). They were rated K3 (n=13) or K4 (n=1) on the Medicare scale and none used an assistive device. The study was approved by the institutional review board for human research and all subjects provided informed consent.

Protocol

Global cognitive function was evaluated using the Modified Mini-Mental Status Exam (3MS) and processing speed and executive function with Trail-Making forms (Trail) A and B while seated. Two cognitive tasks were selected for the dual-task paradigm; serial subtraction by 7 from a 3-digit number and backwards spelling of 5 letter words [27, 28]. Each task was practiced twice before gait assessment.

For gait assessment, subjects walked over an electronic walkway (GAITRite®, length 5.2 m, width 0.6 m). An additional 1.2 m on each end allowed for acceleration/deceleration and recording of a steady state gait. Prior to data collection, subjects made six familiarization passes at normal self-selected speed. They were then instructed to walk at a comfortable pace and simultaneously perform the cognitive task without instructions on prioritization (dual-task gait). Each cognitive task was presented at random in a block of 6 passes. Walks were repeated if the subject stopped on the mat, walked off the side of the mat, had an erratic stepping pattern, or forgot the instructions. After the dual-task conditions, the subjects walked at self-selected normal, slow, and fast speeds, selected freely to ensure natural walking pattern (up to 6 passes each). The normal speed was always collected first, with the order of other two speeds randomized. Demographic and clinical information were collected through an interview and from medical records. All data were collected by the same researcher.

Data processing

Foot fall data from the walkway were processed with a custom program written in MATLAB® (Mathworks Inc., Natick, MA) to calculate instantaneous stride velocity (cm/s), stride length (cm), and stride cadence (strides/min). Stride parameters were calculated from the dominant foot in controls and the prosthetic foot in prosthesis users. Each stride was treated as an individual data point. Trials with at least 5 consecutive steps were included in analysis.

For each subject, the linear regression was used to determine stride length-velocity and cadence-velocity relationships for baseline walking trials at 3 self-selected speeds. To demonstrate goodness of linear fit of stride-length/cadence with velocity at baseline, we reported the coefficient of determination (R^2) for the prosthesis users and controls. Shapriro-Wilk test was used to examine whether regression residuals were normally distributed ($p<0.05$). Using the slope and intercept of the linear equation describing the best fit line, we derived the predicted values for stride length/cadence at the actual stride velocity. The distance (difference) between the actual and predicted values for each data point was calculated (Figure 1). The SD of this difference was adopted as an index of variability in stride length and cadence. Since the method resembles calculation of residuals, we termed it residual standard deviation (RSD). The RSD was calculated for baseline across 3 self-selected speeds to capture the range of natural walking characteristics and for each dual-task condition (subtraction, spelling). The CV (SD/mean) was calculated for stride length and cadence at each self-selected speed (normal, fast, slow) and dual-task conditions (subtraction, spelling).

As subjects are not equally affected by the same tasks [29], the cognitive task (subtraction, spelling) with a more disruptive effect on RSD and CV was selected for each subject for the dual-task analysis (overall 80% task congruence between RSD and CV). The subtraction and spelling tasks were not differently represented between the two groups (Fisher's exact test). Dual-task cost RSD and dual-task cost CV were calculated as the difference between the respective baseline and dual-task conditions for both stride length and cadence (negative sign indicates greater variability during dual-task gait). The slow self-selected speed was chosen as the baseline for deriving dual-task cost CV in prosthesis users because it was comparable to the dual-task speed (78±19 and 82±16 cm/s, respectively). In the control group, the average of the normal CV and slow speed CV was used as the baseline for calculating dual-task cost CV since dual-task speed (112±25) was comparable to the average of these two baseline speeds (114±19 cm/s).

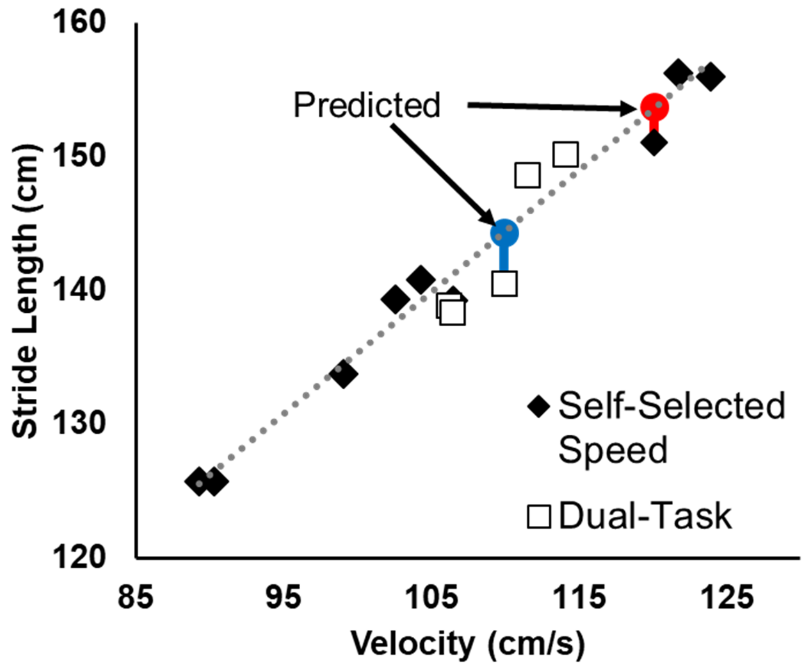


Fig. 1. Calculation of residual standard deviation (RSD). The example shows stride length data points plotted against the respective instantaneous velocity in a prosthesis user (baseline, filled diamonds; dual-task, open squares). Linear regression is fitted first to the baseline data from 3 self-selected speeds. Using the slope and intercept of the linear equation describing the best fit line, the difference between each data point (Actual) and the corresponding point on the best fit line (Predicted) is calculated (baseline, red; dual-task, blue), followed by calculation of the standard deviation of the difference values for each condition.

Statistical analysis

Baseline cognitive performance was compared between prosthesis users and controls (unpaired t-test, $p<0.05$). Velocity, stride length, and cadence were compared for descriptive purposes along with the Pearson's correlation coefficients for stride length-velocity and cadence-velocity relationships (unpaired t-test, $p<0.05$). For construct validity of RSD (aim 1), a 2 x 2 repeated-measure ANOVA was used to compare changes in RSD for stride length and cadence with group (prosthesis users, controls) and condition (baseline, dual-task) as the between and within factors. Hypothesis 1 was tested by significance of the group x condition interaction ($p<0.05$). For concurrent validity (aim 2), the Pearson's correlation coefficient was calculated between dual-task cost RSD and dual-task cost CV for stride length and cadence. Hypothesis 2 was accepted if the correlation was at least moderate (≥ 0.50) [30]. For discriminant validity of dual-task cost RSD (aim 3), the ROC curve was derived for stride length and cadence. Hypothesis 3 was evaluated by the significance of AUC ($p<0.05$). The same was repeated for dual-task cost CV. The shoulder of ROC curve was visually identified and validated by the likelihood ratio. The corresponding cut-off points with sensitivity and specificity were reported. Group data are reported as means and standard deviations (SD). In 4 control subjects (3 stride length and 1 cadence dataset) and 1 prosthesis user (cadence dataset), the linear fit was not adequate based on the runs test. The entire RSD analysis was repeated after replacing the linear with a quadratic model in those 5 cases, but the ROC results remain virtually the same as reported below.

Results

The baseline cognitive performance did not significantly differ between prosthesis users and controls (3MS 96±2 vs. 98±1; Trail A 48±11 vs. 50±6; Trail B 46±10 vs. 53±8, respectively, $p>0.05$). Prosthesis users walked slower, with shorter stride length and at lower cadence than controls at each speed and dual-task condition (Table 1). The R^2 for stride length-velocity and cadence-velocity across 3 self-selected speeds were high for each subject (group means 0.96±0.03 and 0.94±0.04 in prosthesis users, 0.97±0.02 and 0.97±0.01 in controls, respectively). The Shapriro-Wilk test confirmed the normality of residuals in 53 of 54 linear regression analyses, with the only exception of the cadence-velocity residuals in one control subject. The results reported below were not substantially different when the cadence data point for this subject was excluded (not shown).

Table 1
Mean (SD) values for velocity, stride length, and cadence at 3 baseline speeds and under two dual-task conditions for prosthesis users (n=14) and control subjects (n=13). (SUB: serial subtraction; SPL: backwards spelling).

	Fast	Normal	Slow	SUB	SPL
Velocity (cm/s)					
Prosthesis Users	133 (21)*	111 (16)*	82 (16)	78 (20)*	86 (22)*
Control Subjects	161 (23)	134 (21)	94 (21)	112 (24)	113 (24)
Stride length (cm)					
Prosthesis Users	149 (19)	135 (15)	116 (15)	112 (17)*	118 (18)
Control Subjects	158 (15)	145 (15)	121 (16)	131 (17)	131 (17)
Cadence (stride/min)					
Prosthesis Users	53 (5)*	49 (4)*	42 (5)	41 (6)*	43 (6)*
Control Subjects	61 (6)	55 (4)	46 (5)	51 (5)	51 (6)

* Prosthesis Users significantly different from Control Subjects p<0.01

Comparison of RSD between prosthesis users and controls (construct validity of RSD)

The mean RSD values were comparable at baseline, but significantly larger in prosthesis users than controls during dual-task gait for both stride length and cadence (group x condition interaction p=0.0006 and p=0.009, respectively, Table 2). The increase in RSD in prosthesis users was nearly 70% for stride length and 50% for cadence. This confirms the construct validity of RSD (hypothesis 1).

Correlation between dual-task cost RSD and CV (concurrent validity of RSD)

Dual-task cost RSD in prosthesis users was -1.6±1.0 for stride length and -0.54±0.43 for cadence, with the corresponding values for dual-task cost CV of -2.8±2.7 and -2.2±2.6. In controls, these values were lower (stride length RSD -0.11±0.89, CV -1.1±2.0; cadence RSD -0.08±0.39, CV -0.93±1.6). Correlation coefficients between dual-task cost RSD and CV were ≥0.50 for both stride length and cadence (0.52 and 0.50 in prosthesis users, 0.61 and 0.87 in controls, respectively), which confirms the concurrent validity hypothesis.

Table 2
Mean RSD values (SD) for stride length and cadence for prosthesis users and control subjects (group factor in ANOVA) during baseline and dual-task gait (condition factor in ANOVA).

	Baseline	Dual-Task	ANOVA p-values		
			Group	Condition	Interaction
Stride length			0.02	0.0001	**0.0006**
Prosthesis Users	2.5 (0.8)	4.2 (1.6)			
Control Subjects	2.4 (0.7)	2.5 (0.8)			
Cadence			0.03	0.0008	**0.009**
Prosthesis Users	1.0 (0.3)	1.5 (0.6)			
Control Subjects	0.9 (0.3)	1.0 (0.3)			

Table 3
ROC results for discriminative ability of dual-task cost RSD and CV for stride length and cadence. Note highly significant area under the curve (AUC) and better sensitivity and specificity of RSD than CV for the selected cut-off points (significance p-value is in bold).

	AUC (95% CI)	p-value	Cut-off	Likelihood Ratio	Sensitivity	Specificity
Stride Length						
RSD	0.86 (0.73-1.00)	**0.001**	< -0.79	3.71	86%	77%
CV	0.69 (0.49-0.89)	0.089	< -2.53	2.48	57%	77%
Cadence						
RSD	0.81 (0.64-0.98)	**0.007**	< -0.46	9.29	71%	92%
CV	0.65 (0.44-0.86)	0.190	< -0.54	1.55	71%	54%

Comparison of ROC results between RSD and CV (discriminant validity of RSD)

The ROC analysis (Table 3) revealed a significant AUC for dual-task cost RSD for both stride length and cadence (0.863, p=0.001; 0.808, p=0.007, respectively), thereby confirming hypothesis 3. The AUC for dual-task cost CV did not reach significance (stride length 0.692, p=0.089; cadence 0.648, p=0.190). Comparably better discriminative ability of dual-task cost RSD than CV is evident in Figure 2 as the curve offset in the middle of specificity range, along with higher sensitivity or specificity values of the selected cut-off points (Table 3) for both gait parameters.

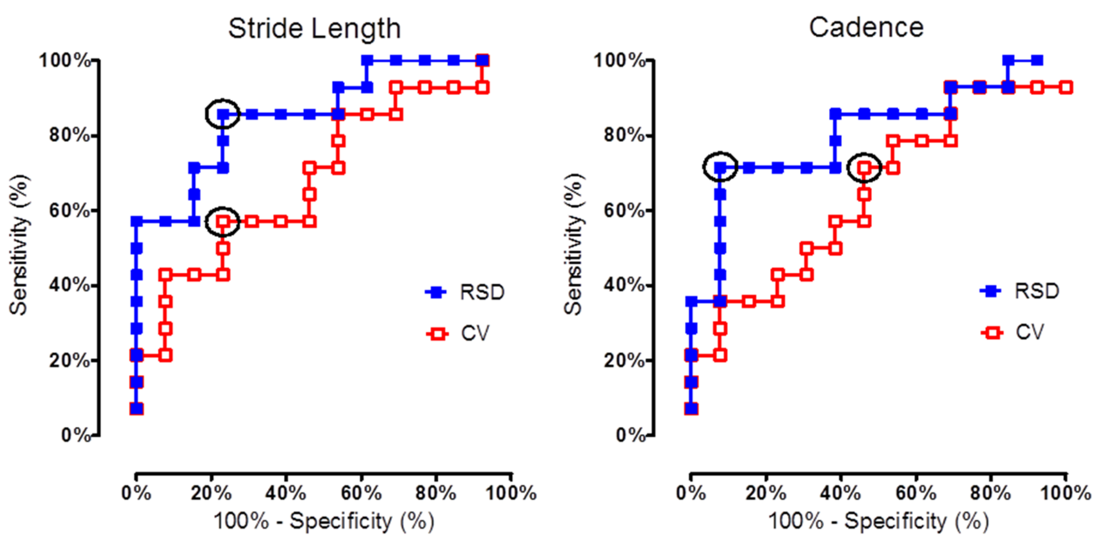


Fig. 2. ROC curves for dual-task cost measured by RSD (blue) and CV (red) for stride length (left) and cadence (right). Note greater area under the curve for RSD than CV for both gait parameters, indicating overall better discriminative ability of RSD. A circle on each curve represents the selected cut-off point for the reported sensitivity and specificity of RSD and CV in discriminating the prosthesis users from controls. While each measure has the same value for stride length specificity and cadence sensitivity note the lack of discriminative power of the complementary value in CV.

Discussion

This study demonstrates the construct, concurrent, and discriminative validity of RSD as a novel index for assessing variability of stride length and cadence during dual-task gait. The three aims were achieved by comparing gait between below-knee prosthesis users and controls walking at different self-selected speeds and while concurrently performing cognitive tasks known to change gait pattern. The construct validity was established by significantly larger increase in RSD from baseline to dual-task gait in prosthesis users compared to controls for both stride length and cadence. The concurrent validity was established by moderate-to-high correlations between the dual-task cost RSD and CV, suggesting that the two measures assess variability in somewhat overlapping but also different ways. The discriminative validity of dual-task cost RSD was confirmed by significant area under ROC curve for both parameters, indicating adequate distinction of prosthesis users from controls. In doing so, dual-task cost RSD outperformed CV. These results have implications for research and clinical practice.

Besides apparent face validity for assessing variability in stride length and cadence, the construct validity of RSD is supported by significant increases in RSD from baseline to dual-task gait condition in the prosthesis users but not controls (Table 2). This was predicted *a priori* based on our previous findings of the disrupted stride length-cadence relationship in below-knee prosthesis users already during natural walking [4]. Intuitively, the addition of a secondary task was likely to further disrupt the gait pattern, which was successfully captured by RSD. Increased variability in prosthesis users could not be attributed to cognitive abilities since cognitive performance at baseline was not different between the two groups. Thus, the possible contributing factors include amputation, impaired sensation, or use and settings of a prosthetic device. All of these, independently or combined, may explain increased cognitive burden during walking. However, two previous studies did not report greater dual-task cost in above-knee prosthesis users compared to controls [14, 15]. The differences may be due to different levels of amputation, cognitive tasks used, or methods employed.

The concurrent validity of dual-task cost RSD is demonstrated by correlations with dual-task cost CV, which were largely in the moderate range, except for being higher for cadence in the controls. Although the reason for the latter is not clear, the overall results suggest that the two indices assess a related but not entirely overlapping construct. Although somewhat predictable given that both measures rely on SD of different but related data points, demonstrating concurrent validity against CV as the criterion was essential for gaining confidence in RSD as a new measure of stride length and cadence variability.

For a measure to be clinically useful, it should be responsive and able to discriminate abnormal from normal gait characteristics. Although in prosthesis users dual-task cost for both stride length and cadence was nominally smaller for RSD (-1.6±1.0, -0.54±0.43) than CV (-2.8±2.7, -2.2±2.6), the proper way to appreciate the responsiveness is to compare the respective standardized response means (mean change/SD of change, in this case dual-task cost mean/SD) [31]. It follows that the responsiveness of dual-task cost RSD (1.6/1.25) is more than one half SD larger than responsiveness of dual-task cost CV (1.0/0.8). In addition, the ROC results confirmed that dual-task cost RSD better differentiates prosthesis users from controls than dual-task cost CV (Table 3, Figure 2). Further studies should examine if RSD is also more sensitive than CV for predicting falls in this and other populations.

The ROC results showed an apparent trade-off in sensitivity and specificity between stride length and cadence. Although this is inherent to ROC analysis (the higher the sensitivity, the lower the specificity, and vice versa), an additional reason for this may be that stride length and cadence are tightly coupled and the changes in one affect the other in the opposite direction. Considering similar AUCs and curve shapes (Table 3, Figure 2), it is to be expected that if one measure (stride length RSD) has greater sensitivity (86%) than specificity (77%), the other related measure (cadence RSD) with comparably lower sensitivity (71%) would yield comparably higher specificity (92%).

The RSD method was used here under the assumption of linear relationship between stride length/cadence and velocity, which has been documented in the vast majority of our cases and as often is the case at typical walking speeds in different populations [1-4]. It should be noted that the RSD approach can also be applied to any non-linear model, because the RSD is a measure of dispersion (SD) of the difference between each observed and predicted point, regardless of how the predicted data point is modelled. However, in this case, the results did not substantially differ when in 5 subjects the RSD values were calculated from the quadratic rather than the linear function with the latter proving slightly less adequate (results not shown).

Limitations

While RSD may also prove to be a useful measure of variability for other parameters related to velocity [19], this validation study was limited to stride length and cadence. Additional gait parameters exhibiting a linear relationship with velocity should be assessed in future studies. Further, this measure was only validated in relatively small samples of below-knee prosthesis users and age-matched controls. Thus, validating this method in above-knee prosthesis users, elderly, and neurological populations is necessary before assuming its broader usefulness. Since this study only assessed the cognitive-to-motor interference aspect of dual-tasking, further analysis of the cognitive performance, including the impact of different tasks, would improve understanding of dual-task behavior in below-knee prosthesis users. Although clinical utility of the RSD method remain unknown, successful validation offers new opportunities for research in this area.

Conclusion

This study validated a new measure of variability in stride length and cadence as the two most robust velocity-dependent gait parameters. The results confirmed the construct, concurrent, and discriminative validity of RSD for measuring dual-task cost in below-knee prosthesis users. The RSD approach may provide more sensitive measures for discriminating between different levels of impairment, monitoring changes in gait performance over time, or examining gait under different conditions variably affecting speed. Further work is needed to determine if this measure of gait variability may be useful for such research and clinical purposes.

References

[1] Egerton T, Danoudis M, Huxham F, Iansek R. Central gait control mechanisms and the stride length - cadence relationship. Gait Posture. 2011; 34:178-82.

[2] Grieve DW, Gear RJ. The relationships between length of stride, step frequency, time of swing and speed of walking for children and adults. Ergonomics. 1966; 5:379-99.

[3] Morris M, Iansek R, Matyas T, Summers J. Abnormalities in stride length-cadence relation in parkinsonian gait. Mov Disord. 1998; 13:61-9.

[4] Howard C, Wallace C, Stokic DS. Stride length-cadence relationship is disrupted in below-knee prosthesis users. Gait Posture. 2013; 38:883-7.

[5] Sawers A, Hafner BJ. Outcomes associated with the use of microprocessor-controlled prosthetic knees among individuals with unilateral transfemoral limb loss: A systematic review. JRRD. 2013; 50:273-314.

[6] Williams RM, Turner AP, Orendurff M, Segal AD, Klute GK, Pecoraro J, et al. Does having a computerized prosthetic knee influence cognitive performance during amputee walking? Arch Phys Med Rehabil. 2006; 87:989-94.

[7] Hafner BJ, Willingham LL, Buell NC, Allyn KJ, Smith DG. Evaluation of function, performance, and preference as transfemoral amputees transition from mechanical to microprocessor control of the prosthetic knee. Arch Phys Med Rehabil. 2007; 88:207-17.

[8] Coffey L, O'Keeffe F, Gallagher P, Desmond D, Lombard-Vance R. Cognitive functioning in persons with lower limb amputations: a review. Disabil Rehabil. 2012; 34:1950-64.

[9] Springer S, Giladi N, Peretz C, Yogev G, Simon ES, Hausdorff JM. Dual-tasking effects on gait variability: the role of aging, falls, and executive function. Mov Disord. 2006; 21:950-7.

[10] Ijmker T, Lamoth CJC. Gait and cognition: The relationship between gait stability and variability with executive function in persons with and without dementia. Gait Posture. 2012; 35:126-30.

[11] Kressig RW, Herrmann FR, Grandjean R, Michel JP, Beauchet O. Gait variability while dual-tasking: fall predictor in older inpatients? Aging Clin Exp Res. 2008; 20:123-30.

[12] Beauchet O, Annweiler C, Allali G, Berrut G, Herrmann FR, Dubost V. Recurrent falls and dual task-related decrease in walking speed: is there a relationship? J Am Geriatr Soc. 2008; 56:1265-9.

[13] Dubost V, Kressig RW, Gonthier R, Herrmann FR, Aminian K, Najafi B, et al. Relationships between dual-task related changes in stride velocity and stride time variability in healthy older adults. Hum Mov Sci. 2006; 25:372-82.

[14] Morgan SJ, Hafner BJ, Kelly VE. The effects of a concurrent task on walking in persons with transfemoral amputation compared to persons without limb loss. Prosthet Orthot Int. 2016; 40:490-6.

[15] Lamoth CJ, Ainsworth E, Polomski W, Houdijk H. Variability and stability analysis of walking of transfemoral amputees. Med Eng Phys. 2010; 32:1009-14.

[16] Amboni M, Barone P, Hausdorff JM. Cognitive contributions to gait and falls: evidence and implications. Mov Disord. 2013; 28:1520-33.

[17] Lord S, Howe T, Greenland J, Simpson L, Rochester L. Gait variability in older adults: A structured review of testing protocol and clinimetric properties. Gait Posture. 2011; 34:443-50.

[18] Lord S, Galna B, Verghese J, Coleman S, Burn D, Rochester L. Independent domains of gait in older adults and associated motor and nonmotor attributes: validation of a factor analysis approach. J Gerontol A Biol Sci Med Sci. 2013; 68:820-7.

[19] Chisholm AE, Makepeace S, Inness EL, Perry SD, McIlroy WE, Mansfield A. Spatial-temporal gait variability poststroke: variations in measurement and implications for measuring change. Arch Phys Med Rehabil. 2014; 95:1335-41.

[20] Beauchet O, Annweiler C, Lecordroch Y, Allali G, Dubost V, Herrmann F, et al. Walking speed-related changes in stride time variability: effects of decreased speed. J Neuroeng Rehabil. 2009; 6:32.

[21] Yogev-Seligmann G, Rotem-Galili Y, Dickstein R, Giladi N, Hausdorff JM. Effects of explicit prioritization on dual task walking in patients with Parkinson's disease. Gait Posture. 2012; 35:641-6.

[22] Wrightson JG, Ross EZ, Smeeton NJ. The Effect of Cognitive-Task Type and Walking Speed on Dual-Task Gait in Healthy Adults. Motor Control. 2016; 20:109-21.

[23] Gates DH, Dingwell JB, Scott SJ, Sinitski EH, Wilken JM. Gait characteristics of individuals with transtibial amputations walking on a destabilizing rock surface. Gait Posture. 2012; 36:33-9.

[24] Hordacre B, Bradnam LV, Barr C, Patritti BL, Crotty M. Ipsilateral corticomotor excitability is associated with increased gait variability in unilateral transtibial amputees. Eur J Neurosci. 2014; 40:2454-62.

[25] Nordin E, Moe-Nilssen R, Ramnemark A, Lundin-Olsson L. Changes in step-width during dual-task walking predicts falls. Gait Posture. 2010; 32:92-7.

[26] Singh R, Hunter J, Philip A, Tyson S. Gender differences in amputation outcome. Disabil Rehabil. 2008; 30:122-5.

[27] Yogev G, Giladi N, Peretz C, Springer S, Simon ES, Hausdorff JM. Dual tasking, gait rhythmicity, and Parkinson's disease: which aspects of gait are attention demanding? Eur J Neurosci. 2005; 22:1248-56.

[28] Hollman JH, Kovash FM, Kubik JJ, Linbo RA. Age-related differences in spatiotemporal markers of gait stability during dual task walking. Gait Posture. 2007; 26:113-9.

[29] Patel P, Lamar M, Bhatt T. Effect of type of cognitive task and walking speed on cognitive-motor interference during dual-task walking. Neuroscience. 2014; 260:140-8.

[30] Hinkle DE, Wiersma W, Jurs SG. Applied statistics for the behavioral sciences. Boston: Houghton Mifflin; 2003.

[31] Husted JA, Cook RJ, Farewell VT, Gladman DD. Methods for assessing responsiveness: a critical review and recommendations. J Clin Epidemiol. 2000; 53:459-68.

CHAPTER 8

PROSTHESIS USERS HAVE INCREASED GAIT VARIABILITY WHILE WALKING DURING CHALLENGING GAIT CONDITIONS AND DUAL-TASKING

Abstract

Residual standard deviation (RSD) utilizes the linear relationships between stride length, cadence, and velocity to calculate variability of stride length and cadence in a manner that allows for differences in gait speed across trials. RSD was used to assess changes in gait variability in 10 below-knee prosthesis users and 12 control subjects during challenging gait conditions: a narrow walkway and walking with a tray, with and without performing a cognitive task (dual-task). Subjects performed the dual-tasks without and then with instruction to prioritize the cognitive task performance. Without the cognitive task, the narrow walkway increased stride length and cadence variability in prosthesis users more than in control subjects (Group x Walk interaction $p \leq 0.002$). Walking with the tray did not impact variability in either group (Walk $p > 0.4$). Dual-tasking without instruction on prioritization did not increase stride length or cadence variability in either group during normal or narrow walking (Task $p > 0.1$). However, performing a cognitive task while walking with a tray increased stride length and cadence variability in prosthesis users (Walk x Task interaction $p \leq 0.032$) and stride length variability in control subjects (Task $p = 0.018$). When instructed to prioritize the cognitive task the prosthesis users did not exhibit a change in variability or cognitive performance ($p > 0.2$). However, control subjects did improve cognitive performance (Instruction $p = 0.040$) and had an increase in cadence variability (Instruction $p = 0.022$) during the narrow walking condition. The results suggest that prosthesis users are more disrupted by

a narrow walking constraint than control subjects but this challenge does not increase cognitive burden. Walking while balancing a tray, however, does not impact gait variability but may increase the cognitive burden of walking. These results suggest prosthesis users are more disrupted by highly demanding gait tasks but that the specific characteristics of the challenge influence the relative impact on neuromotor and cognitive processes.

Introduction

The performance of daily activities requires the allocation of resources among motor, sensory, and cognitive systems [1]. When the demands of an activity are low, such as quiet walking in a flat well-lit area, resource demands are low. However, a more challenging gait condition, such as a narrow walkway, may increase demand [2, 3]. Further, many daily activities require multitasking and may require resources to be allocated to multiple systems or for a single system to be engaged in more than one task. When demand increases, resources may become limited and may not able to meet the needs of all tasks, which can degrade performance of one or more of the tasks [1].

A dual-task paradigm is often used to simulate multitasking in a research setting. Specifically, dual-tasking is the concurrent performance of two tasks with independent goals and outcomes [2]. Dual-tasking often involves standing or walking while performing a cognitive task, such as serial subtraction. Dual-task performance has been used to identify unstable gait patterns and fall risk [4-9] but has also been used to probe how cognitive resources are allocated in task performance to better understand neural processing [10-12]. For example, studies finding decrements in the performance of one or both tasks suggest that the decline is due to competition for attentional resources by the different systems needed to perform the different tasks. This model for neural processing is referred to as the cross-domain competition model [10] or the central-capacity sharing model [12]. However, there are other studies that find improved performance on one or both tasks. Improved performance in the presence of dual-tasking supports a prioritization model [10, 11] or increased recruitment of previously unengaged resources, increased level of alertness [13, 14]. Understanding resource allocation can help with developing strategies to reduce the risk of multi-tasking induced falls in at risk populations and study designs aimed at capturing specific risks.

In order to best examine resource allocation while dual-tasking, cognitive resources must be stressed to provoke a measurable change in performance. This can be achieved by utilizing challenging gait conditions in conjunction with a dual-task evaluation. Challenging single-task gait conditions increase the demand of the walking without changing the goal [2]. A narrow walkway was used to impose a physical constraint; walking while carrying a tray was used to increase task complexity [2]. The flexibility or the ability to consciously adjust resource allocation can be probed by instructing subjects on specific task prioritization [11, 15]. This methodology was successfully used to evaluate dual-task strategy in unilateral below-knee prosthesis users and non-amputee control subjects during a standing dual-task evaluation [14]. In this study, the same approach was used to further evaluate dual-task performance during challenging gait conditions in these groups.

The linear coupling between stride length and cadence has been used as a measure of coordinated neurocontrol of gait [16-18]. Chapter 4 showed that prosthesis users had less coordination between stride length and cadence compared to control subjects during a simple walking task [19]. Increased resource demand imposed by a challenging gait task may further disrupt the stride length-cadence relationship.

Dual-task studies using simple and challenging gait conditions in above-knee prosthesis users with standard gait measures have not identified an effect of dual-tasking on most aspects of gait, including step time variability [20, 21]. However, dual-task evaluation of below-knee prosthesis users with the residual standard deviation (RSD) method, a novel method of variability analysis, identified greater dual-task impact in prosthesis users than control subjects while performing a normal walking task (Chapter 7) [22]. In this study, the impact of dual-tasking on gait under challenging conditions was evaluated utilizing the residual standard deviation (RSD) method due to its demonstrated utility in the below-knee prosthesis user population.

The first aim was to evaluate the impact of challenging single-task gait conditions on coordination and variability of stride length and cadence. Hypothesis 1 was that the challenging conditions will decrease coordination and increase variability in both groups but more so in prosthesis users than control subjects. The second aim was to evaluate the impact of performing a cognitive task while walking in challenging gait conditions on stride length and cadence variability. Hypothesis 2 was that dual-tasking would have a greater effect on prosthesis users stride length and cadence variability in the challenging gait conditions than the normal condition. Furthermore we evaluated the flexibility of resource allocation while walking in a challenging gait condition by instructing subjects to focus on performance of the cognitive task.

Methods

Participants

A group of below-knee prosthesis users and non-amputee control subjects who had previously participated in a standing dual-task study [14] returned between two weeks to one month later to participate in a gait dual-task evaluation. The inclusion criteria for the prosthesis users were ≥ 1 year since amputation; age 18–80 years; comfortable socket fit; no known balance, neurological, or other health problems that limit daily activities; and able to safely walk 10 m at different velocities, as verified by a certified prosthetist. The study was approved by the institutional review board for human research and all subjects signed the informed consent.

The sample included 12 control subjects (mean ± SD age 47±14 years, 15±2 years of education, BMI 30.8±7 kg/m^2, 5 (42%) men) and 10 prosthesis users (mean ± SD age 47±13 years, 13±3 years of education, BMI 29.7±6 kg/m^2, 7 (70%) men). The amputation occurred 7±7 years earlier (range 1-18) due to trauma (n = 5) or vascular disease (n = 5). All prosthesis users were rated K3 on the Medicare scale and none used an assistive device.

Protocol

Footfall placement and timing for each walk was recorded using a 6-m electronic walkway (Zeno Walkway®, ProtoKinetics, Havertown, PA) with an additional 1.2 m on each end to allow for acceleration/deceleration. The walking conditions used for analysis were unrestrained walking (normal), walking with a narrow base of support (narrow), and walking while carrying a tray with a cup filled with ping pong balls (tray). The width of the narrow path for each subject was 50% of their anterior superior iliac spine width plus their shoe width [23]. The path was designated by covering the edges of the walkway with black poster board, which provided a strong contrast against the walkway. During the tray task subjects used both hands to carry a typical cafeteria tray. A cup filled with ping pong balls was placed in the center of the tray to encourage subjects to keep the tray level while walking. The walking conditions were presented in random order. During each walking condition subjects were instructed to make 4 passes at a normal speed followed by passes at fast and slow self-selected speeds presented in a random order.

Two cognitive tasks were selected for the dual-task paradigm; serial subtraction by 7 from a 3-digit number and a verbal fluency task (listing words starting with a specific letter). The most difficult letters for verbal fluency (J, K, Q, U, X, Y, Z) were excluded from this task [24]. Each task was practiced while seated to ensure comprehension. For comparison between groups, the subtraction task was performed 2 times for 30 s and the verbal fluency F-A-S test (FAS) was performed once for 60 s. The number of correct responses was documented, and the verbal responses were also recorded to confirm response accuracy.

After a seated break, each of the three walking conditions were repeated at the subjects' self-selected speed while performing the cognitive tasks. Subjects made 2 passes for each task in each walking condition. The walking conditions and cognitive tasks were presented in random order. In the first set of trials, subjects were given no instruction on task prioritization. In the second set, subjects were asked to focus on the cognitive task and increase the number of correct responses by at least 50% over their noted average in the first presentation. In each set, an additional subtraction task was given at random as a distractor (subtracting 6 or 8, data not included). The performance on each cognitive task was documented and recorded. The average number of correct responses for each task in each walking condition was calculated for analysis.

Data processing

Data from the electronic walkway were processed with a custom program written in MATLAB® (Mathworks Inc., Natick, MA) to calculate instantaneous stride velocity (cm/s), stride length (cm), and stride cadence (strides/min). Each variable was calculated from the dominant foot in control subjects and the prosthetic foot in prosthesis users. Each stride was treated as an individual data point.

For each single-task walking condition, the linear regression across the three walking speeds was calculated between stride length and cadence to evaluate the stride length-cadence relationship. The goodness of fit was evaluated using the coefficient of determination (R^2) of the regression. See Chapter 4 [19] for detailed description of the analysis.

Stride length and cadence variability was calculated using the RSD method for single- and dual-task walking. The linear regression between stride length and velocity and cadence and velocity was calculated for each single-task walking condition across the 3 self-selected speeds. Using the formulas for the best fit line, the predicted value for stride length/cadence was calculated from the instantaneous velocity for the respective walking condition. The difference between the actual and predicted values for each point was calculated and the variability of stride length/cadence was measured as the standard deviation of the differences. The RSD was measured for each single-task walking condition and the respective dual-task conditions (subtraction, verbal fluency). See Chapter 7 [22] for detailed description of the analysis.

Due to personal aptitude, the same task may not have equal interfering effect in all subjects. As in previous studies [14, 22], the cognitive task (subtraction or verbal fluency) with a more disruptive effect across the 3 walking conditions was identified in each subject and selected for the dual-task analysis in order to ensure all subjects were stressed by dual-tasking. The distribution of the subtraction and verbal fluency tasks was not different between the two groups (Fisher's exact test $p=1$).

Dual-task cost was calculated for each walking condition as the difference in RSD variability between the single-task and selected dual-task for the no prioritization and prioritization conditions (negative sign indicates increased variability during dual-task walking).

Statistical analysis

The seated cognitive task performance was compared between the prosthesis users and control subjects (unpaired t-test, $p \leq 0.05$). As the two challenging walking conditions (narrow and tray) induced perturbation through different modalities, physical constraint vs. complex task, the conditions were evaluated in separate analyses. To test whether the more challenging walking conditions had greater impact on prosthesis users than control subjects the stride length-cadence relationship and the single-task stride length and cadence RSD (hypothesis 1) was evaluated with a 2x2 mixed ANOVA with Group (prosthesis users, control subjects) as the between-subjects factor and Walk (normal, narrow; normal, tray) as the within-subjects factor. The differential response of the two groups to the challenging walking tasks was determined by a Group interaction ($p \leq 0.05$). When main effects and interactions were present, adjusted t-tests ($p \leq 0.025$) were performed on within subject factors to facilitate interpretation of the results.

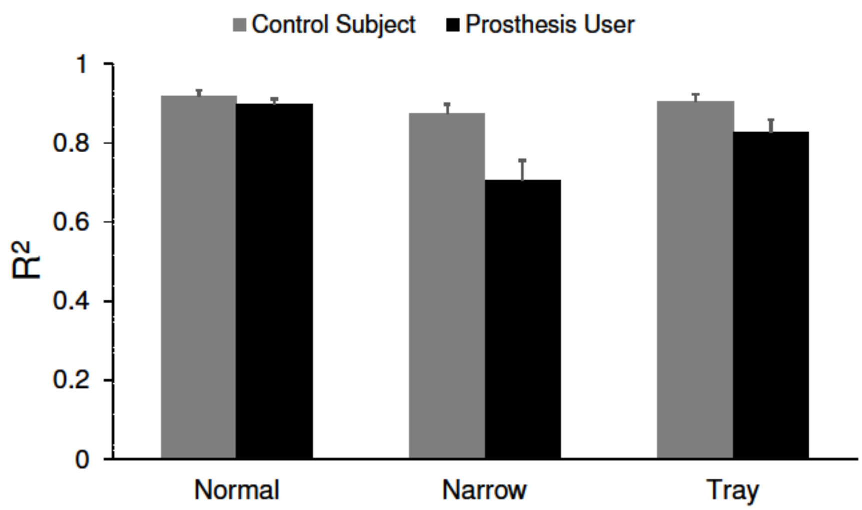
The impact of performing a cognitive dual-task on stride length and cadence RSD (hypothesis 2) was evaluated within each group using a 2x2 repeated measure ANOVA with Walk (normal, narrow; normal, tray) and Task (single, dual) as factors. An increase in dual-task impact on variability during the challenging gait conditions was determined by the Walk x Task interaction ($p \leq 0.05$). The result of instructing subjects to prioritize the cognitive task performance was evaluated using the RSD dual-task cost within each group using a 2x2 repeated measure ANOVA with Walk (normal, narrow; normal, tray) and Instruction (no prioritization, prioritization) as factors. The impact of the prioritization instruction on cognitive task performance was further evaluated using the same analysis. IBM SPSS Statistics 23 (IBM Corp., Armonk, NY) was used for statistical analysis.

Results

There was no difference between groups for seated performance of the cognitive tasks. The total number of correct responses during the FAS test were 39±3 (mean ± SE) words for control subjects and 43±4 words for prosthesis users ($p=0.4$). The average number of correct subtraction responses were 6.6±1 for control subjects and 5.3±1 for prosthesis users ($p=0.4$).

Single-task stride length-cadence relationship (hypothesis 1)

During single-task walking, the narrow walkway (Walk-narrow $p<0.001$; paired t-test: control $p=0.023$; prosthesis users $p=0.003$) and carrying the tray (Walk-tray $p=0.007$) disrupted the stride length-cadence relationship in both groups (Figure 1). However, the narrow walking condition resulted in a greater disruption in prosthesis users than control subjects (Group x Walk-narrow interaction p=0.005). These results confirm hypothesis 1 for the narrow but not the tray walking condition.

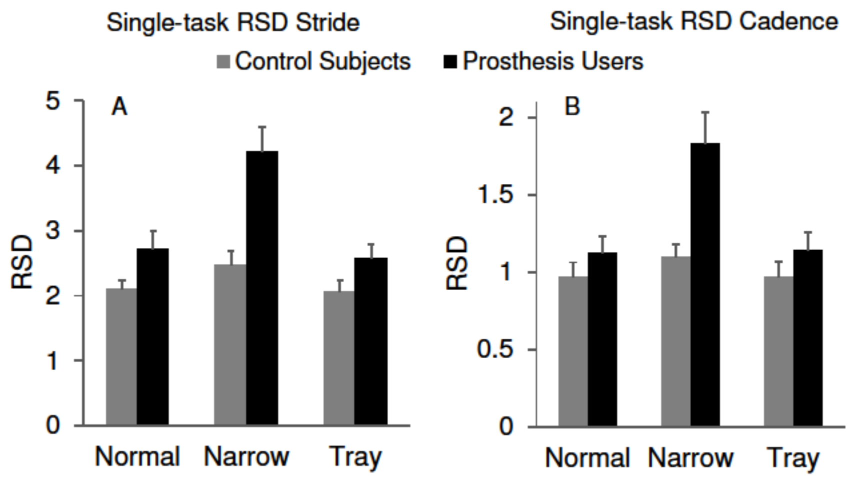


ANOVA p-values	Narrow	Tray
Group	**0.006**	0.064
Walk	**<0.001**	**0.007**
Group x Walk	**0.005**	0.071

Fig. 1. R^2 values of the stride length-cadence relationship for control subjects (gray) and prosthesis users (black) during each single-task walking condition. The narrow walking condition reduced the R^2 value in both groups but caused greater disruption in the linear relationship in prosthesis users than control subjects. Carrying a tray had similar disruption to the stride length-cadence relationship in both groups.

Single-task stride length and cadence variability (hypothesis 1)

Narrow walking also significantly increased stride length and cadence variability, as indicated by an increase in RSD, in prosthesis users greater than in control subjects (Group x Walk-narrow interaction p=0.001 and p=0.003, respectively). The narrow walking condition did not increased stride length or cadence variability in control subjects (paired t-test: stride length p=0.040; cadence p=0.1). Walking while carrying a tray did not increase stride length or cadence variability in either group (Walk-tray main effects and interactions p>0.5). For stride length there was a main effect of group for the analysis of both walking conditions (Group-narrow p=0.002; Group-tray p=0.035), suggesting that prosthesis users had higher RSD variability across all walking conditions. These results also confirm hypothesis 1 for stride length and cadence variability for the narrow but not the tray walking condition. Figure 2 shows the single-task RSD results.



ANOVA p-values	Stride Length		Cadence	
	Narrow	Tray	Narrow	Tray
Group	**0.002**	**0.035**	**0.011**	0.207
Walk	**<0.001**	0.523	**<0.001**	0.942
Group x Walk	**0.001**	0.679	**0.003**	0.922

Fig. 2. RSD during each single-task walking condition for each control subjects (gray) and prosthesis users (black). Prosthesis users had higher stride length variability across walking conditions; however, the narrow walking condition resulted in the greatest difference between groups. The narrow walking condition also increased stride length variability in control subjects. Prosthesis users' cadence variability was only higher than control subjects in the narrow walking condition. Walking while carrying a tray did not increase variability over the normal walking condition in either group.

Single-task vs. no-prioritization dual-task stride length and cadence variability (hypothesis 2)

There was no main effect of Task or Walk x Task interaction for either group for the dual-task narrow walk analysis on stride length and cadence variability (Walk-narrow main effects and interactions p>0.1; Figure 3 A/B). However, while carrying a tray, prosthesis users had a significant Walk x Task interaction for both stride length and cadence, showing higher variability while carrying a tray and performing a cognitive task (Walk-tray x Task interaction p=0.050 and p=0.032, respectively; Figure 3 C/D). This confirms hypothesis 2 for the tray but not the narrow walking condition. Control subjects also had a main effect of task for stride length variability in the tray dual-task analysis only (Task-tray p=0.019). Considering no significant effects were observed in the analysis of the narrow walks, this suggests that performing a cognitive task while carrying a tray also increased stride length variability in control subjects.

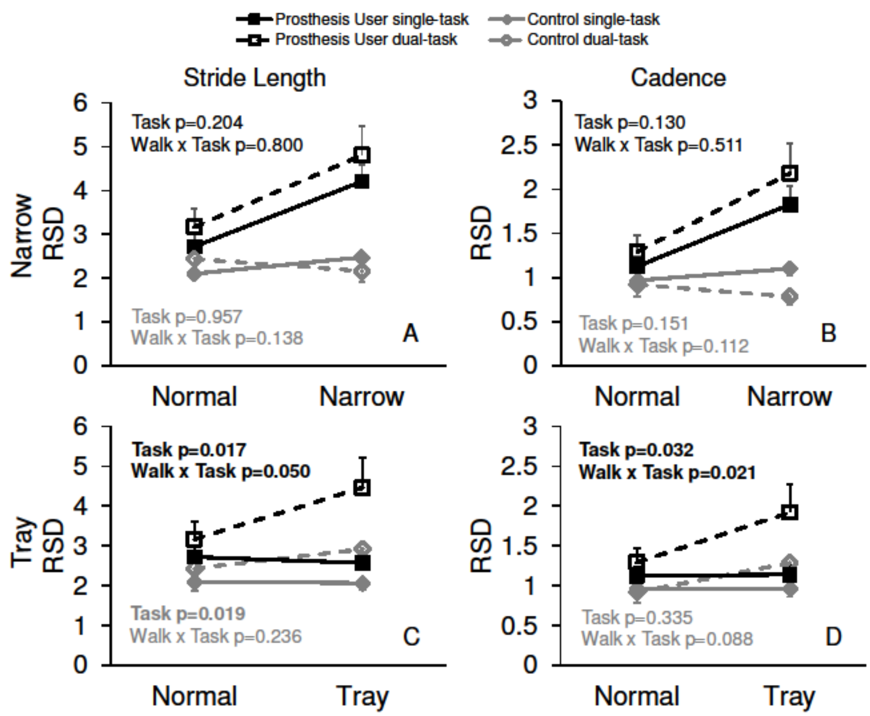


Fig. 3. RSD during single-task (solid line) and no-prioritization dual-task (dashed line) walking during the challenging walking conditions. Dual-tasking during the narrow walking condition did not increase stride length (A) or cadence (B) variability in either group. However, dual-tasking while carrying the tray increased both stride length (C) and cadence (D) variability in prosthesis users (black) and stride length (C) variability in control subjects (gray).

No-prioritization vs. prioritization dual-task cost stride length and cadence variability

The instruction to prioritize the performance of the cognitive task did not impact the dual-task cost for prosthesis users across walking conditions (Instruction main effect and interactions p>0.2, Figure 4). However, the prioritization instruction did result in increased dual-task cost in cadence during the narrow walking condition (Instruction-narrow p=0.022; Figure 4b).

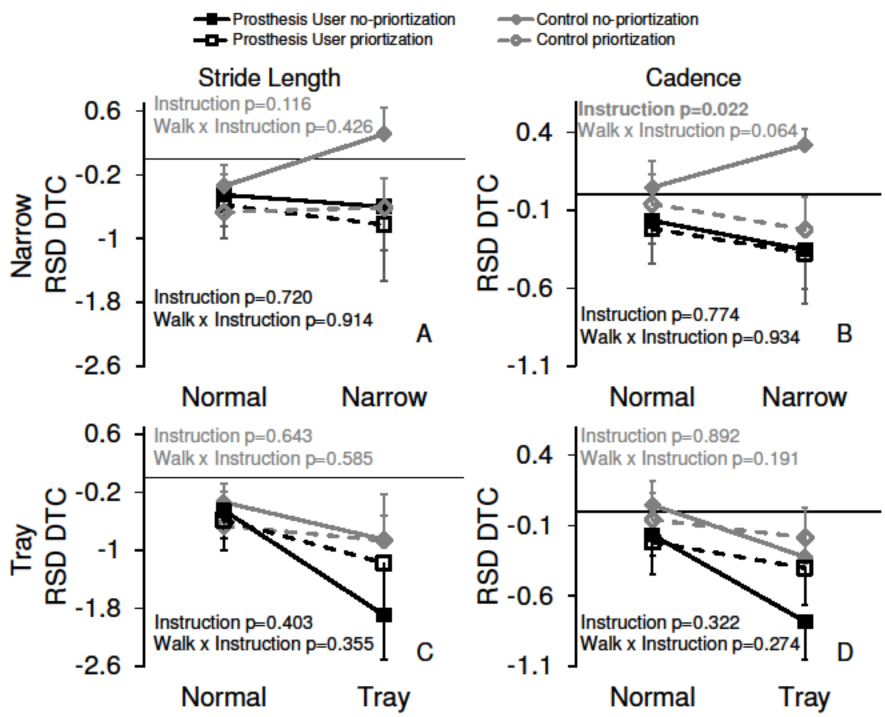


Fig. 4. RSD dual-task cost for the no-prioritization instruction (solid line) and prioritization instruction (dashed line) walking during the challenging walking conditions. Overall the instruction to prioritize the cognitive task did not impact stride length or cadence variability. However, control subjects (gray) did have an increase in cadence variability during the narrow walking condition (B).

Dual-task cognitive performance

Both prosthesis users and control subjects successfully performed the cognitive tasks while walking. Across all conditions and tasks the average number of correct responses was between 3 and 4 while the average number of incorrect responses was less than 1. The walking condition did not significantly impact the average number of correct responses in either group (Walk $p>0.3$). The instruction to prioritize the cognitive task also did not impact the number of correct responses in prosthesis users (Instruction main effect and interactions $p>0.6$). However, control subjects did increase their average number of correct responses by 1 when instructed to prioritize the cognitive task during the normal vs. narrow walking analysis (Instruction-narrow $p=0.040$).

Discussion

This study demonstrated the impact of challenging walking tasks on the control of gait in prosthesis users and control subjects. The nature of the gait challenge affects the type of stresses experienced by locomotion control systems. The constraint of a narrow walkway disrupted gait coordination and variability in both groups but did not appear to greatly increase the cognitive burden. However, the complexity of carrying a tray while walking did not disrupt gait control until it was combined with performance of a cognitive task. While overall, prosthesis users had less coordinated gait and greater variability, the narrow walking condition increased the difference. The impact of the cognitive task on the tray condition also caused a more consistent disruption in prosthesis users. For most conditions, the instruction to prioritize the cognitive task did not result in a change in performance for either the walking condition or the cognitive task. However, during the narrow walking condition the control subjects did improve cognitive task performance; this corresponded with an increase in cadence variability. These results indicate that goal prioritization, even under single-task conditions, can affect cognitive resource allocation and this may be an important consideration in the design of future studies.

In both groups, the narrow walking task decreased coordination between stride length and cadence while increasing variability, but dual-tasking did not exacerbate that difference. This finding is similar to those from Kelly et al. who found a main effect of dual-task performance on gait speed for usual and narrow walking but no interaction in young health adults [25]. Although the physical demands of the narrow walking condition required alterations in gait mechanics, the constraints imposed by the condition did not pose enough risk to warrant additional cognitive resource allocation. As such, while the narrow task might be physically challenging it is not in-fact demanding of substantial focus. This may also explain findings from Morgan et al. that reported no greater dual-task impact in above-knee prosthesis users while walking on a compliant surface [21]. Alternatively, the lack of dual-task impact in the narrow walking condition could be due to prioritization of the walking task, as suggested by Kelly et al [25]. However, in this study, there was no effect of walking condition on the number of correct responses in either group. This argues against a change in prioritization between walking conditions. It was only when control subjects deliberately utilized more resources to improve cognitive task performance that gait variability increased. This relationship between greater cognitive resource allocation and increased gait variability does suggest that there was competition between resources for overall dual-task performance and that subjects may have the flexibility to direct resource allocation based on their goal.

Walking while carrying a tray had an opposite impact on single- and dual-task gait compared to the narrow walking condition. The tray task did not significantly impact single-task walking but had a significant dual-task effect in both groups. While it can be considered a complex single-task rather than a motor dual-task [2], carrying a tray may have increased the importance of the walking goal to subjects. Thus, it appears subjects may have dedicated more resources to walking during the single-task condition, therefore limiting the impact on gait coordination and variability. These results support the notion of flexible resource allocation and that the goal of the individual may play an important role in how resources are used [11]. In this case, the goal of the walking task was to maintain a stable walking patterning. However, when prosthesis users performed standing goal-oriented tasks, if best achievement of the goal conflicted with the most stable stance (standing on the intact side), prosthesis users often choose to compromise stability (standing on the prosthetic side) [26]. This further highlights the goal-directed nature of resource allocation.

Performing a cognitive task while carrying a tray increased stride length and cadence variability in prosthesis users and stride length variability in control subjects. This further supports the theory that maintenance of the gait pattern in the single-task condition required additional resources than normal walking. The greater dual-task impact in the tray condition shows a competition for resources and supports a resource sharing and competition model [10, 12]. The results also indicate that the gait task was not prioritized.

While the groups had similar patterns of performance, when the walking condition or dual-task did impact gait, it was stronger in the prosthesis users. This supports other findings of impaired gait mechanics in prosthesis users [19] and increased cognitive burden of using a prosthesis [14, 22]. The greater single-task impact of the narrow walking condition suggests that prosthesis users are less capable of adapting to the medial-lateral disturbance imposed by the narrow walkway. Also, while control subjects did have a significant dual-task effect during the tray condition, the effect was more consistent in prosthesis users. This suggests that use of a prosthesis may increase the overall cognitive burden of walking and may make prosthesis users less able to respond to increased demands in everyday life. These findings may represent an increased risk of instability and falls in prosthesis users [27-30] .

Limitations

This study utilized a small non-homogenous sample. A larger sample would have given the statistical tests greater power and perhaps highlighted significance in observed patterns that did not show significant differences in this study. While the diverse sample gives a general picture of below-knee prosthesis user behavior, it may limit clinical interpretation for specific patient groups. Future studies should evaluate the impact of dual-tasking with challenging gait conditions in larger and more homogenous groups. The order of the instructions should also be considered. The instruction to prioritize the cognitive task was always presented after the no-prioritization condition, which increases the risk of a learning effect. However, as the instruction only impacted the results in control subjects during the narrow walking condition, rather than across groups and conditions, and resulted in a differential effect on gait and cognitive performance, this argues against the presence of a strong learning effect. It should also be noted that the subjects had previously participated in a similar dual-task protocol for posture analysis. However, subjects commented on the continued difficulty of performing the cognitive task.

Conclusion

The results show that when assessing the impact of experimental challenges on gait, the nature of the challenge imposes on subjects should be considered. When the challenge imposes a high cost of gait deviations, subjects may dedicate enough resources such that a change in performance may not be apparent until the resources are depleted, resulting in competition. When the challenge does not remarkably increase the cost of gait deviations, gait may be more likely to be impacted as subjects do not dedicate additional resources to the task. Thus, not all challenging tasks may evoke additional use of cognitive resources. The results also show that prosthesis users behave similarity to control subjects in response to different demands and goals, however the effects are more pronounced and suggest that prosthesis users may be a greater risk of falls. Prosthetic devices that improve medial-lateral control and stability or reduce the cognitive burden of using a prosthesis may reduce demands for cognitive resources, which could help to decrease fall risk in below-knee prosthesis users.

References

[1] Woollacott M, Shumway-Cook A. Attention and the control of posture and gait: a review of an emerging area of research. Gait Posture. 2002; 16:1-14.

[2] McIsaac TL, Lamberg EM, Muratori LM. Building a framework for a dual task taxonomy. Biomed Res Int. 2015; 2015:1-10.

[3] Lindenberger U, Marsiske M, Baltes PB. Memorizing while walking: increase in dual-task costs from young adulthood to old age. Psychol Aging. 2000; 15:417.

[4] Hausdorff JM, Schweiger A, Herman T, Yogev-Seligmann G, Giladi N. Dual-task decrements in gait: contributing factors among healthy older adults. J Geront A. 2008; 63:1335-43.

[5] Springer S, Giladi N, Peretz C, Yogev G, Simon ES, Hausdorff JM. Dual-tasking effects on gait variability: the role of aging, falls, and executive function. Mov Disord. 2006; 21:950-7.

[6] Sample RB, Jackson K, Kinney AL, Diestelkamp WS, Reinert SS, Bigelow KE. Manual and Cognitive Dual Tasks Contribute to Fall-Risk Differentiation in Posturography Measures. J Appl Biomech. 2016; 32:541-7.

[7] McCulloch KL, Buxton E, Hackney J, Lowers S. Balance, attention, and dual-task performance during walking after brain injury: associations with falls history. J Head Trauma Rehabil. 2010; 25:155-63.

[8] Muir-Hunter SW, Wittwer JE. Dual-task testing to predict falls in community-dwelling older adults: a systematic review. Physiotherapy. 2016; 102:29-40.

[9] Nordin E, Moe-Nilssen R, Ramnemark A, Lundin-Olsson L. Changes in step-width during dual-task walking predicts falls. Gait Posture. 2010; 32:92-7.

[10] Lacour M, Bernard-Demanze L, Dumitrescu M. Posture control, aging, and attention resources: Models and posture-analysis methods. Clin Neurophysiol. 2008; 38:411-21.

[11] Yogev-Seligmann G, Hausdorff JM, Giladi N. Do we always prioritize balance when walking? Towards an integrated model of task prioritization. Mov Disord. 2012; 27:765-70.

[12] Beurskens R, Steinberg F, Antoniewicz F, Wolff W, Granacher U. Neural correlates of dual-task walking: Effects of cognitive versus motor interference in young adults. Neural Plast. 2016; 2016:1-9.

[13] Bonnet CT, Baudry S. Active vision task and postural control in healthy, young adults: Synergy and probably not duality. Gait Posture. 2016; 48:57-63.

[14] Howard CL, Perry B, Chow JW, Wallace C, Stokic DS. Increased alertness, better than posture prioritization, explains dual-task performance in prosthesis users and controls under increasing postural and cognitive challenge. Exp Brain Res. 2017; Epub: Aug 31; 10.1007/s00221-017-5077-2.

[15] Yogev-Seligmann G, Rotem-Galili Y, Mirelman A, Dickstein R, Giladi N, Hausdorff JM. How does explicit prioritization alter walking during dual-task performance? Effects of age and sex on gait speed and variability. Phys Ther. 2010; 90:177-86.

[16] Egerton T, Danoudis M, Huxham F, Iansek R. Central gait control mechanisms and the stride length - cadence relationship. Gait Posture. 2011; 34:178-82.

[17] Zijlstra W, Rutgers AWF, Hof AL, Van Weerden TW. Voluntary and involuntary adaptation of walking to temporal and spatial constraints. Gait Posture. 1995; 3:13-8.

[18] Morris M, Iansek R, Matyas T, Summers J. Abnormalities in stride length-cadence relation in parkinsonian gait. Mov Disord. 1998; 13:61-9.

[19] Howard C, Wallace C, Stokic DS. Stride length-cadence relationship is disrupted in below-knee prosthesis users. Gait Posture. 2013; 38:883-7.

[20] Morgan SJ, Hafner BJ, Kelly VE. The effects of a concurrent task on walking in persons with transfemoral amputation compared to persons without limb loss. Prosthet Orthot Int. 2016; 40:490-6.

[21] Morgan SJ, Hafner BJ, Kelly VE. Dual-task walking over a compliant foam surface: A comparison of people with transfemoral amputation and controls. Gait Posture. 2017; 58:41-5.

[22] Howard CL, Wallace C, Abbas J, Stokic DS. Residual standard deviation: Validation of a new measure of dual-task cost in below-knee prosthesis users. Gait Posture. 2017; 51:91-6.

[23] Gimmon Y, Jacob G, Lenoble-Hoskovec C, Büla C, Melzer I. Relative and absolute reliability of the clinical version of the Narrow Path Walking Test (NPWT) under single and dual task conditions. Arch Gerontol Geriatr. 2013; 57:92-9.

[24] Borkowski JG, Benton AL, Spreen O. Word fluency and brain damage. Neuropsychologia. 1967; 5:135-40.

[25] Kelly VE, Eusterbrock AJ, Shumway-Cook A. Factors influencing dynamic prioritization during dual-task walking in healthy young adults. Gait Posture. 2013; 37:131-4.

[26] Howard C, Wallace C, Stokic DS. Lower limb preference on goal-oriented tasks in unilateral prosthesis users. Gait Posture. 2012; 36:249-53.

[27] Schrager MA, Kelly VE, Price R, Ferrucci L, Shumway-Cook A. The effects of age on medio-lateral stability during normal and narrow base walking. Gait Posture. 2008; 28:466-71.

[28] Cho Bl, Scarpace D, Alexander NB. Tests of stepping as indicators of mobility, balance, and fall risk in balance-impaired older adults. J Am Geriatr Soc. 2004; 52:1168-73.

[29] Etemadi Y. Dual task cost of cognition is related to fall risk in patients with multiple sclerosis: a prospective study. Clin Rehabil. 2017; 31:278-84.

[30] Hollman JH, Kovash FM, Kubik JJ, Linbo RA. Age-related differences in spatiotemporal markers of gait stability during dual task walking. Gait Posture. 2007; 26:113-9.

CHAPTER 9

DISCUSSION

This work sought to develop new protocols to assess posture and mobility control in lower-limb prosthesis users that provide better utility than existing, commonly employed methods. Using the framework of cognitive resource allocation, maintenance of stability while standing and walking were assessed during normal and difficult conditions, with and without performance of a concurrent task. In addition to the experimental protocols designed to evoke changes in cognitive resource allocation, spectral analysis of posture identified previously unobserved changes in the use of the somatosensory system. Furthermore, the residual standard deviation (RSD) method, a novel method to measure gait variability, was developed and demonstrated higher sensitivity and specificity than traditional variability measures. Overall, the results illustrate the importance of task goals and prioritization in resource allocation. These findings addressed the specific aims presented in the introduction and provide new insight into prosthesis users' cognitive resource allocation while performing competing tasks.

In evaluating specific aim 1 (Chapter 3), it was found that prosthesis users often utilize their prosthetic leg for balance when performing goal-oriented standing tasks that require unilateral use of the lower limbs. This was counter to the hypothesized behavior expected due to typical physical therapy training [1] and studies showing high reliance on the intact side for stability during standing and walking [2-4]. The findings of this study suggest that motivation to achieve certain goals may take higher priority over maintenance of stability. Chapters 5-8, which evaluated dual-task performance while standing and walking, also found that prosthesis users were more likely than non-amputee control subjects to have disruption to gait or posture stability while concurrently performing a cognitive task, confirming the first hypotheses of aims 3 and 4. Together, these studies further support a tendency towards prioritization of goals separate from posture or gait control in prosthesis users and may point to reasons for increased fall risk in the population [5].

When evaluating prosthesis users in challenging standing or walking conditions, for most conditions, the subjects continued to prioritize performance of the cognitive task despite the increased risk to stability. This was counter to hypothesis 2 of both aims 3 and 4, postulating that the increased challenge would cause subjects to allocate more resources to maintenance of stability. In addition to further confirming the dual-task behavior of prosthesis users, the dual-task performance during the challenging conditions also supported theories of cognitive resource competition [6, 7].

The differential performance between the narrow and tray walking conditions (Chapter 8) provides strong evidence of the influence of cognitive resource demand and competition on motor behavior. When evaluating single-task walking, the narrow condition resulted in an increase in stride length and cadence variability along with lower gait coordination, but walking while carrying a tray did not. When the conditions were coupled with performance of a cognitive task, the tray condition increased gait variability while there was no change in the narrow condition. The inability to maintain a consistent gait pattern while dual-tasking and carrying a tray suggests that the tray condition was more cognitively demanding than the other walking conditions. Without dual-tasking, the demand of the tray walking condition would not have been apparent and may have been interpreted as an un-challenging walking condition. However, the dual-task results suggest that the tray condition was demanding but that cognitive resources were used to maintain a consistent gait pattern. Thus, the changes in the gait pattern, interpreted as representing the challenge of the walking condition, were not apparent until the resources were further stressed beyond the limits of the postural reserve. Similarly, the narrow walking condition results could also be misinterpreted. In being described as a challenging walking condition, its use is expected to make dual-tasking more difficult and evoke a stronger response. Taken alone, the narrow walking results would be interpreted as showing no dual-task impact and suggestive of no greater cognitive burden in prosthesis users. However, when considered in regard to the tray condition and in light of cognitive resource theories, the disruption to the gait pattern in the single-task walking condition suggest that the narrow condition did not prompt subjects to allocate greater resources towards stability. Thus, they were available for performance of the dual-task conditions. The differential findings of the walking conditions are important for study designs and interpretations and further highlight the importance of goals in subject's allocation of resources.

Further support for cognitive resource reorganization to accommodate competition while dual-tasking comes from the spectral analysis of the center of pressure signal while standing (Chapter 6). While dual-tasking, both control subjects and prosthesis users had a decrease in the frequency band associated with postural control adjustments driven by visual control. This suggests that more resources were allocated to other control systems while dual-tasking. This finding confirms the third hypothesis of aim 3.

The RSD method proved to be more effective at evaluating dual-task cost than traditional measures of variability during a normal walking task, while also being effective in the evaluation of gait variability during challenging walking conditions, confirming hypothesis 3 of aim 4. While this analysis method was developed to address issues of sensitivity in studies evaluating prosthesis users, the RSD could be applied to other populations and it is not limited to dual-task analysis. Additionally, while the analysis was only applied to stride length and cadence variability, the mathematical principle (Appendix C) could be applied to other variables whose linear relationship affects variability calculations.

The principle of RSD emerged from the evaluation stride length-cadence relationship. Not only did this analysis provide the framework for a novel variability calculation, but also highlighted the utility of evaluating the coupling between stride/step length and cadence as a comprehensive measure of gait quality; as it was effective in distinguishing prosthetic gait from control subjects, confirming hypotheses 1 and 2 from aim 2. From a statistical standpoint, the comprehensive measure reduces the risk of statistical error by providing a single outcome effectively representing three variables, stride length, cadence, and velocity, across a range of self-selected speeds. It also avoids pitfalls of other protocols that may inadvertently alter subjects' natural walking pattern, such as walking on a treadmill or restricting subjects to a specific walking speed.

Future work

The methods developed and utilized in the research could also be used to identify differences between different prosthetic populations, such as fallers and non-fallers, or to distinguish between different prosthetic devices. Utilizing the receiver operating characteristic curve, potential cut-off values between normal and abnormal gait measured by the stride length-cadence relationships and RSD were identified. In future work, these values could help to establish clinical classifications of gait deviations or provide targets for achieving “normal gait”. Further, the high sensitivity and specificity of these measures show promise for distinguishing between different groups or devices. Future work should include larger samples sizes of more homogenous groups to confirm cut-off values and evaluate the effectiveness of the classification.

The use of spectral analysis on the center of pressure signal identified a smaller contribution from the prosthetic side in the frequency band associated with the somatosensory system. As emerging research in lower-limb devices is working towards the incorporation of sensory feedback systems [8-11], the ability to specifically measure the impact on somatosensory control could prove useful in the evaluation of these devices. Further work is needed to evaluate this measurement of the somatosensory contribution and assess the responsiveness of spectral analysis to lower limb sensory changes. In addition to continued evaluation of lower-limb prosthesis users, the effectiveness of the measure could be assessed in persons with peripheral neuropathy.

In addition to further evaluation of the analysis techniques, future work may also benefit from the insights into dual-task methodology that were developed. Through pilot testing it was identified that subjects do not have equal aptitude for the cognitive tasks utilized in the dual-task analysis which can translate to dual-task performance. It is easy to consider that a mathematician may not require as many resources to perform a serial subtraction task as a writer, while the writer may excel at generating lists of words. By utilizing diverse tasks and analyzing only the most disruptive, all subjects were more likely to be challenged by the dual-task paradigm, providing a more uniform representation of performance. This novel method proved useful across all the dual-task studies. Also, this work stresses the importance of considering not just the challenge of the experimental manipulation used to stress subjects but how that challenge affects the goal. For example, while the narrow walking condition in this work did not appear to increase subjects' cognitive burden, a raised narrow walkway might evoke a different response as the goal of maintaining stability might receive greater weight. Future work should not only consider the type of challenging condition but also continue to evaluate differences between different types of conditions to better understand how the challenge impacts resource allocation.

Conclusion

These findings address many of the central issues highlighted by Sawers et al. [12] by focusing on the prosthesis user-device interaction rather than on specific component features. The results support the notion that utilizing a prosthesis imposes substantial cognitive demands while also suggesting that prosthesis users may place higher prioritization on achieving goals rather than maintenance of stability. This work also increased the knowledge on the utility of dual-task protocols, particularly regarding the importance of goal prioritization, and provided a new measure of gait variability that proved to be more effective at identifying gait changes in prosthesis users than traditional measures. Future studies utilizing these methods may provide new information to clinicians and researchers on prosthesis users' behavior, motor control strategies, and fall risk. The analysis methods used, particularly the stride length-cadence relationship, residual standard deviation, and spectral analysis of the center of pressure signal may provide unique and sensitive methods to assess differences in prosthetic devices.

References

[1] Pasquina PF, Cooper RA. Care of the Combat Amputee. Textbooks of Military Medicine. Washington, DC: Office of the Surgeon General; 2010. p. 820.

[2] Ku PX, Abu Osman NA, Wan Abas WA. Balance control in lower extremity amputees during quiet standing: A systematic review. Gait Posture. 2014; 39:672-82.

[3] Nolan L, Wit A, Dudzinski K, Lees A, Lake M, Wychowanski M. Adjustments in gait symmetry with walking speed in trans-femoral and trans-tibial amputees. Gait Posture. 2003; 17:142-51.

[4] Gailey R, Allen K, Castles J, Kucharik J, Roeder M. Review of secondary physical conditions associated with lower-limb amputation and long-term prosthesis use. J Rehabil Res Dev. 2008; 45:15-29.

[5] Hunter SW, Batchelor F, Hill KD, Hill A-M, Mackintosh S, Payne M. Risk Factors for Falls in People With a Lower Limb Amputation: A Systematic Review. PMR. 2017; 9:170-80.

[6] Woollacott M, Shumway-Cook A. Attention and the control of posture and gait: a review of an emerging area of research. Gait Posture. 2002; 16:1-14.

[7] Lacour M, Bernard-Demanze L, Dumitrescu M. Posture control, aging, and attention resources: Models and posture-analysis methods. Clin Neurophysiol. 2008; 38:411-21.

[8] Fan RE, Culjat MO, King CH, Franco ML, Boryk R, Bisley JW, et al. A haptic feedback system for lower-limb prostheses. IEEE Trans Neural Syst Rehabil Eng. 2008; 16:270-7.

[9] Rusaw D, Hagberg K, Nolan L, Ramstrand N. Can vibratory feedback be used to improve postural stability in persons with transtibial limb loss? J Rehabil Res Dev. 2012; 49:1239-54.

[10] Marasco P, Hebert J, Makhlin A, Shell C, Forero J. Physiologically Relevant Prosthetic Limb Movement Feedback for Upper and Lower Extremity Amputees. The Cleveland Clinic Foundation Cleveland United States; 2016.

[11] Crea S, Cipriani C, Donati M, Carrozza MC, Vitiello N. Providing time-discrete gait information by wearable feedback apparatus for lower-limb amputees: usability and functional validation. IEEE Trans Neural Syst Rehabil Eng. 2015; 23:250-7.

[12] Sawers A, Hahn ME, Kelly VE, Czerniecki J, Kartin D. Beyond componentry: How principles of motor learning can enhance locomotor rehabilitation of individuals with lower limb loss—A review. J Rehabil Res Dev. 2012; 49:1431-42.

WORKS CITED

Aggashyan RV, Gurfinkel VS, Mamasakhlisov GV, Elner AM. Changes in spectral and correlation characteristics of human stabilograms at muscle afferentation disturbance. Agressologie. 1973; 14:5-9.

Aldridge JM, Sturdy JT, Wilken JM. Stair ascent kinematics and kinetics with a powered lower leg system following transtibial amputation. Gait Posture. 2012; 36:291-5.

Amboni M, Barone P, Hausdorff JM. Cognitive contributions to gait and falls: evidence and implications. Mov Disord. 2013; 28:1520-33.

Andriacchi TP, Ogle JA, Galante JO. Walking speed as a basis for normal and abnormal gait measurments. J Biomech. 1977; 10:261-8.

Au S, Berniker M, Herr H. Powered ankle-foot prosthesis to assist level-ground and stair-descent gaits. Neural Netw. 2008; 21:654-66.

Barak Y, Wagenaar RC, Holt KG. Gait characteristics of elderly people with a history of falls: a dynamic approach. Physical Therapy. 2006; 86:1501-10.

Beauchet O, Dubost V, Aminian K, Gonthier R, Kressig RW. Dual-task-related gait changes in the elderly: does the type of cognitive task matter? J Mot Behav. 2005; 37:259-64.

Beauchet O, Annweiler C, Allali G, Berrut G, Dubost V. Dual task-related changes in gait performance in older adults: a new way of predicting recurrent falls? J Am Geriatr Soc. 2008; 56:181-2.

Beauchet O, Annweiler C, Allali G, Berrut G, Herrmann FR, Dubost V. Recurrent falls and dual task-related decrease in walking speed: is there a relationship? J Am Geriatr Soc. 2008; 56:1265-9.

Beauchet O, Annweiler C, Lecordroch Y, Allali G, Dubost V, Herrmann F, et al. Walking speed-related changes in stride time variability: effects of decreased speed. J Neuroeng Rehabil. 2009; 6:32.

Bensel CK, Dzendolet E. Power spectral density analysis of the standing sway of males. Atten Percept Psychophys. 1968; 4:285-8.

Berge J, Czerniecki J, Klute GK. Efficacy of shock-absorbing versus rigid pylons for impact reduction in transtibial amputees based on laboratory, field, and outcome metrics. J Rehabil Res Dev. 2005; 42:795-808.

Bernard-Demanze L, Dumitrescu M, Jimeno P, Borel L, Lacour M. Age-related changes in posture control are differentially affected by postural and cognitive task complexity. Curr Aging Sci. 2009; 2:139-49.

Beurskens R, Steinberg F, Antoniewicz F, Wolff W, Granacher U. Neural correlates of dual-task walking: Effects of cognitive versus motor interference in young adults. Neural Plast. 2016; 2016:1-9.

Bisson EJ, Lajoie Y, Bilodeau M. The influence of age and surface compliance on changes in postural control and attention due to ankle neuromuscular fatigue. Exp Brain Res. 2014; 232:837-45.

Bloem BR, Grimbergen YA, van Dijk JG, Munneke M. The "posture second" strategy: a review of wrong priorities in Parkinson's disease. J Neurol Sci. 2006; 248:196-204.

Bonnet CT, Baudry S. Active vision task and postural control in healthy, young adults: Synergy and probably not duality. Gait Posture. 2016; 48:57-63.

Borkowski JG, Benton AL, Spreen O. Word fluency and brain damage. Neuropsychologia. 1967; 5:135-40.

Borrenpohl D, Kaluf B, Major MJ. Survey of U.S. Practitioners on the Validity of the Medicare Functional Classification Level System and Utility of Clinical Outcome Measures for Aiding K-Level Assignment. Arch Phys Med Rehabil. 2016; 97:1053-63.

Bronstein AM, Brandt T, Woollacott MH, Nutt JG. Clinical Disorders of Balance, Posture and Gait. 2nd ed. London: Arnold Publishers; 2004.

Buckley JG, O'Driscoll D, Bennett SJ. Postural sway and active balance performance in highly active lower-limb amputees. Am J Phys Med Rehabil. 2002; 81:13-20.

Callisaya ML, Blizzard L, McGinley JL, Srikanth VK. Risk of falls in older people during fast-walking - The TASCOG study. Gait Posture. 2012; 36:510-5.

Chagdes JR, Rietdyk S, Haddad JM, Zelaznik HN, Raman A, Rhea CK, et al. Multiple timescales in postural dynamics associated with vision and a secondary task are revealed by wavelet analysis. Exp Brain Res. 2009; 197:297-310.

Chen R, Corwell B, Yaseen Z, Hallett M, Cohen LG. Mechanisms of cortical reorganization in lower-limb amputess. J Neurosci. 1998; 18:3443-50.

Chisholm AE, Makepeace S, Inness EL, Perry SD, McIlroy WE, Mansfield A. Spatial-temporal gait variability poststroke: variations in measurement and implications for measuring change. Arch Phys Med Rehabil. 2014; 95:1335-41.

Cho Bl, Scarpace D, Alexander NB. Tests of stepping as indicators of mobility, balance, and fall risk in balance-impaired older adults. J Am Geriatr Soc. 2004; 52:1168-73.

Coffey L, O'Keeffe F, Gallagher P, Desmond D, Lombard-Vance R. Cognitive functioning in persons with lower limb amputations: a review. Disabil Rehabil. 2012; 34:1950-64.

Collins JJ, De Luca CJ. Open-loop and closed-loop control of posture: a random-walk analysis of center-of-pressure trajectories. Exp Brain Res. 1993; 95:308-18.

Collins JJ, De Luca CJ. Random walking during quiet standing. Phys Rev Lett. 1994; 73:764-7.

Collins JJ, De Luca CJ, Burrows A, Lipsitz LA. Age-related changes in open-loop and closed-loop postural control mechanisms. Exp Brain Res. 1995; 104:480-92.

Crea S, Cipriani C, Donati M, Carrozza MC, Vitiello N. Providing time-discrete gait information by wearable feedback apparatus for lower-limb amputees: usability and functional validation. IEEE Trans Neural Syst Rehabil Eng. 2015; 23:250-7.

Delussu AS, Brunelli S, Paradisi F, Iosa M, Pellegrini R, Zenardi D, et al. Assessment of the effects of carbon fiber and bionic foot during overground and treadmill walking in transtibial amputees. Gait Posture. 2013; 38:876-82.

Dessery Y, Barbier F, Gillet C, Corbeil P. Does lower limb preference influence gait initation? Gait Posture. 2011; 33:550-5.

Diener HC, Dichgans J, Bruzek W, Selinka H. Stabilization of human posture during induced oscillations of the body. Exp Brain Res. 1982; 45:126-32.

Diener HC, Dichgans J, Bacher M, Gompf B. Quantification of postural sway in normals and patients with cerebellar diseases. Electroencephalogr Clin Neurophysiol. 1984; 57:134-42.

Diener HC, Dichgans J, Guschlbauer B, Mau H. The significance of proprioception on postural stabilization as assessed by ischemia. Brain Res. 1984; 296:103-9.

Du Chatinier K, Molen NH, Rozendal RH. Step length, step frequency and temporal factors of the stride in normal human walking. Anatomy. 1970; 73:214-27.

Dubost V, Kressig RW, Gonthier R, Herrmann FR, Aminian K, Najafi B, et al. Relationships between dual-task related changes in stride velocity and stride time variability in healthy older adults. Hum Mov Sci. 2006; 25:372-82.

Eberly VJ, Mulroy SJ, Gronley JK, Perry J, Yule WJ, Burnfield JM. Impact of a stance phase microprocessor-controlled knee prosthesis on level walking in lower functioning individuals with a transfemoral amputation. Prosthet Orthot Int. 2014; 38:447-55.

Egerton T, Danoudis M, Huxham F, Iansek R. Central gait control mechanisms and the stride length - cadence relationship. Gait Posture. 2011; 34:178-82.

Elias LJ, Bryden MP. Footedness is a better predictor of language lateralisation than handedness. Laterality. 1998; 3:41-51.

Elias LJ, Bryden MP, Bulman-Fleming MB. Footedness is a better predictor than is handedness of emotional lateralization. Neuropsychologia. 1998; 36:37-43.

Etemadi Y. Dual task cost of cognition is related to fall risk in patients with multiple sclerosis: a prospective study. Clin Rehabil. 2017; 31:278-84.

Fan RE, Culjat MO, King CH, Franco ML, Boryk R, Bisley JW, et al. A haptic feedback system for lower-limb prostheses. IEEE Trans Neural Syst Rehabil Eng. 2008; 16:270-7.

Fok P, Farrell M, McMeeken J. Prioritizing gait in dual-task conditions in people with Parkinson's. Hum Mov Sci. 2010; 29:831-42.

Fritz NE, Cheek FM, Nichols-Larsen DS. Motor-Cognitive Dual-Task Training in Persons With Neurologic Disorders: A Systematic Review. J Neurol Phys Ther. 2015; 39:142-53.

Fujita H, Kasubuchi K, Wakata S, Hiyamizu M, Morioka S. Role of the Frontal Cortex in Standing Postural Sway Tasks While Dual-Tasking: A Functional Near-Infrared Spectroscopy Study Examining Working Memory Capacity. Biomed Res Int. 2016; Epub: Feb 3; 10.1155/2016/7053867.

Gailey R, Allen K, Castles J, Kucharik J, Roeder M. Review of secondary physical conditions associated with lower-limb amputation and long-term prosthesis use. J Rehabil Res Dev. 2008; 45:15-29.

Gates DH, Dingwell JB, Scott SJ, Sinitski EH, Wilken JM. Gait characteristics of individuals with transtibial amputations walking on a destabilizing rock surface. Gait Posture. 2012; 36:33-9.

Geil MD. Recommendations for Research on Microprocessor Knees. J Prosthet Orthot. 2013; 25:P76-P9.

Geurts AC, Mulder TW, Nienhuis B, Rijken RA. Dual-task assessment of reorganization of postural control in persons with lower limb amputation. Arch Phys Med Rehabil. 1991; 72:1059-64.

Geurts AC, Mulder TW, Nienhuis B, Rijken RA. Postural reorganization following lower limb amputation. Possible motor and sensory determinants of recovery. Scand J Rehabil Med. 1992; 24:83-90.

Geurts AC, Mulder TH. Attention demands in balance recovery following lower limb amputation. J Mot Behav. 1994; 26:162-70.

Gholizadeh H, Abu Osman NA, Eshraghi A, Ali S. Transfemoral prosthesis suspension systems: a systematic review of the literature. Am J Phys Med Rehabil. 2014; 93:809-23.

Gholizadeh H, Abu Osman NA, Eshraghi A, Ali S, Razak NA. Transtibial prosthesis suspension systems: systematic review of literature. Clin Biomech. 2014; 29:87-97.

Ghulyan V, Paolino M, Lopez C, Dumitrescu M, Lacour M. A new translational platform for evaluating aging or pathology-related postural disorders. Acta Otolaryngol. 2005; 125:607-17.

Gimmon Y, Jacob G, Lenoble-Hoskovec C, Büla C, Melzer I. Relative and absolute reliability of the clinical version of the Narrow Path Walking Test (NPWT) under single and dual task conditions. Arch Gerontol Geriatr. 2013; 57:92-9.

Grieve DW, Gear RJ. The relationships between length of stride, step frequency, time of swing and speed of walking for children and adults. Ergonomics. 1966; 5:379-99.

Grouios G. Footedness as a potential factor that contributes to the causation of corn and callus formation in lower extremities of physically active individuals. The Foot. 2005; 15:154-62.

Hafner BJ, Sanders JE, Czerniecki J, Fergason J. Energy storage and return prostheses: does patient perception correlate with biomechanical analysis? Clin Biomech. 2002; 17:325-44.

Hafner BJ. Clinical prescription and use of prosthetic foot and ankle mechanisms: a review of the literature. J Prosthet Orthot. 2005; 17:S5-S11.

Hafner BJ. Overview of outcome measures for the assessment of prosthetic foot and ankle componets. J Prosthet Orthot. 2006; 18:105.

Hafner BJ, Willingham LL, Buell NC, Allyn KJ, Smith DG. Evaluation of function, performance, and preference as transfemoral amputees transition from mechanical to microprocessor control of the prosthetic knee. Arch Phys Med Rehabil. 2007; 88:207-17.

Hafner BJ, Smith DG. Differences in function and safety between Medicare Functional Classification Level-2 and-3 transfemoral amputees and influence of prosthetic knee joint control. J Rehabil Res Dev. 2009; 46:417-33.

Hamacher D, Brennicke M, Behrendt T, Alt P, Torpel A, Schega L. Motor-cognitive dual-tasking under hypoxia. Exp Brain Res. 2017; Epub: Jul 18; 10.1007/s00221-017-5036-y.

Hansen AH. Scientific Methods to determine functional performance of prosthetic ankle-foot systems. J Prosthet Orthot. 2005; 17:S23-S9.

Hausdorff JM, Schweiger A, Herman T, Yogev-Seligmann G, Giladi N. Dual-task decrements in gait: contributing factors among healthy older adults. J Geront A. 2008; 63:1335-43.

Hermodsson Y, Ekdahl C, Persson B, Roxendal G. Standing balance in trans-tibial amputees following vascular disease or trauma: a comparative study with healthy subjects. Prosthet Orthot Int. 1994; 18:150-8.

Highsmith MJ, Schulz BW, Hart-Hughes S, Latlief GA, Phillips SL. Differences in the spatiotemporal parameters of transtibial and transfemoral amputee gait. J Prosthet Orthot. 2010; 22:26-30

Hinkle DE, Wiersma W, Jurs SG. Applied statistics for the behavioral sciences. Boston: Houghton Mifflin; 2003.

Hofstad C, Linde H, Limbeek J, Postema K. Prescription of prosthetic ankle-foot mechanisms after lower limb amputation. Cochrane Database Syst Rev. 2004; Epub: Feb 20; 10.1002/14651858.CD003978.pub2.

Hollman JH, Kovash FM, Kubik JJ, Linbo RA. Age-related differences in spatiotemporal markers of gait stability during dual task walking. Gait Posture. 2007; 26:113-9.

Hollman JH, Youdas JW, Lanzino DJ. Gender differences in dual task gait performance in older adults. Am J Mens Health. 2011; 5:11-7.

Hordacre B, Bradnam LV, Barr C, Patritti BL, Crotty M. Ipsilateral corticomotor excitability is associated with increased gait variability in unilateral transtibial amputees. Eur J Neurosci. 2014; 40:2454-62.

Houdijk H, Pollmann E, Groenewold M, Wiggerts H, Polomski W. The energy cost for the step-to-step transition in amputee walking. Gait Posture. 2009; 30:35-40.

Howard C, Wallace C, Stokic DS. Lower limb preference on goal-oriented tasks in unilateral prosthesis users. Gait Posture. 2012; 36:249-53.

Howard C, Wallace C, Stokic DS. Stride length-cadence relationship is disrupted in below-knee prosthesis users. Gait Posture. 2013; 38:883-7.

Howard CL, Perry B, Chow JW, Wallace C, Stokic DS. Increased alertness, better than posture prioritization, explains dual-task performance in prosthesis users and controls under increasing postural and cognitive challenge. Exp Brain Res. 2017; Epub: Aug 31; 10.1007/s00221-017-5077-2.

Howard CL, Wallace C, Abbas J, Stokic DS. Residual standard deviation: Validation of a new measure of dual-task cost in below-knee prosthesis users. Gait Posture. 2017; 51:91-6.

Huang CY, Lin LL, Hwang IS. Age-Related Differences in Reorganization of Functional Connectivity for a Dual Task with Increasing Postural Destabilization. Front Aging Neurosci. 2017; 9:96.

Huang HJ, Mercer VS. Dual-task methodology: applications in studies of cognitive and motor performance in adults and children. Pediatr Phys Ther. 2001; 13:133-40.

Hunter SW, Batchelor F, Hill KD, Hill A-M, Mackintosh S, Payne M. Risk Factors for Falls in People With a Lower Limb Amputation: A Systematic Review. PMR. 2017; 9:170-80.

Husted JA, Cook RJ, Farewell VT, Gladman DD. Methods for assessing responsiveness: a critical review and recommendations. J Clin Epidemiol. 2000; 53:459-68.

Ijmker T, Lamoth CJC. Gait and cognition: The relationship between gait stability and variability with executive function in persons with and without dementia. Gait Posture. 2012; 35:126-30.

In F, Kim S. Introduction to Wavelet Theory in Finance : A Wavelet Multiscale Approach. Singapore, UNITED STATES: World Scientific Publishing Company; 2012.

Isakov E, Mizrahi J, Ring H, Susak Z, Hakim N. Standing sway and weight-bearing distribution in people with below-knee amputations. Arch Phys Med Rehabil. 1992; 73:174-8.

Isakov E, Burger H, Krajnik J, Gregoric M, Marincek C. Influence of speed on gait parameters and on symmetry in transtibial amputees. Prosthet Orthot Int. 1996; 20:153-8.

Kalaycioglu C, Kara C, Atbasoglu C, Nalcaci E. Aspects of foot preference: differential relationships of skilled and unskilled foot movements with motor asymmetry. Laterality. 2008; 13:124-42.

Kelly VE, Schrager MA, Price R, Ferrucci L, Shumway-Cook A. Age-associated effects of a concurrent cognitive task on gait speed and stability during narrow-base walking. J Geront A. 2008; 63:1329-34.

Kelly VE, Eusterbrock AJ, Shumway-Cook A. Factors influencing dynamic prioritization during dual-task walking in healthy young adults. Gait Posture. 2013; 37:131-4.

Kirchner M, Schubert P, Schmidtbleicher D, Haas C. Evaluation of the temporal structure of postural sway fluctuations based on a comprehensive set of analysis tools. Phys A. 2012; 391:4692-703.

Klodd E, Hansen AH, Fatone S, Edwards M. Effects of prosthetic foot forefoot flexibility on oxygen cost and subjective preference rankngs of unilateral transtibial prosthesis users. J Rehabil Res Dev. 2010; 47:543-52.

Klute GK, Kallfelz CF, Czerniecki J. Mechanical properties of prosthetic limbs:Adapting to the patient. J Rehabil Res Dev. 2001; 38:299-307.

Klute GK, Berge JS, Orendurff MS, Williams RM, Czerniecki JM. Prosthetic intervention effects on activity of lower-extremity amputees. Arch Phys Med Rehabil. 2006; 87:717-22.

Klute GK, Kantor C, Darrouzet C, Wild H, Wilkinson S, Iveljic S, et al. Lower-limb amputee needs assessments using multistakeholder focus-group approach. J Rehabil Res Dev. 2009; 46:293-304.

Klute GK, Glaister BC, Berge JS. Prosthetic liners for lower limb amputees: a review of the literature. Prosthet Orthot Int. 2010; 34:146-53.

Kressig RW, Herrmann FR, Grandjean R, Michel JP, Beauchet O. Gait variability while dual-tasking: fall predictor in older inpatients? Aging Clin Exp Res. 2008; 20:123-30.

Ku PX, Abu Osman NA, Wan Abas WA. Balance control in lower extremity amputees during quiet standing: A systematic review. Gait Posture. 2014; 39:672-82.

Lacour M, Bernard-Demanze L, Dumitrescu M. Posture control, aging, and attention resources: Models and posture-analysis methods. Clin Neurophysiol. 2008; 38:411-21.

Lamberg EM, Muratori LM. Cell phones change the way we walk. Gait Posture. 2012; 35:688-90.

Lamoth CJ, Ainsworth E, Polomski W, Houdijk H. Variability and stability analysis of walking of transfemoral amputees. Med Eng Phys. 2010; 32:1009-14.

Legro MW, Reiber G, del Aguila M, Ajax MJ, Boone DA, Larsen JA, et al. Issues of importance reported by persons with lower limb amputations and prostheses. J Rehabil Res Dev. 1999; 36:155-63.

Lindenberger U, Marsiske M, Baltes PB. Memorizing while walking: increase in dual-task costs from young adulthood to old age. Psychol Aging. 2000; 15:417.

Liston MB, Bergmann JH, Keating N, Green DA, Pavlou M. Postural prioritization is differentially altered in healthy older compared to younger adults during visual and auditory coded spatial multitasking. Gait Posture. 2014; 39:198-204.

Lord S, Howe T, Greenland J, Simpson L, Rochester L. Gait variability in older adults: A structured review of testing protocol and clinimetric properties. Gait Posture. 2011; 34:443-50.

Lord S, Galna B, Verghese J, Coleman S, Burn D, Rochester L. Independent domains of gait in older adults and associated motor and nonmotor attributes: validation of a factor analysis approach. J Geront A. 2013; 68:820-7.

Maatar D, Fournier R, Lachiri Z, Nait-Ali A. Discrete wavelet and modified PCA decompositions for postural stability analysis in biometric applications. J Biomed Sci Eng. 2011; 4:543.

Marasco P, Hebert J, Makhlin A, Shell C, Forero J. Physiologically Relevant Prosthetic Limb Movement Feedback for Upper and Lower Extremity Amputees. The Cleveland Clinic Foundation Cleveland United States; 2016.

Martinez-Ramirez A, Lecumberri P, Gomez M, Izquierdo M. Wavelet analysis based on time-frequency information discriminate chronic ankle instability. Clin Biomech. 2010; 25:256-64.

Martinez-Ramirez A, Lecumberri P, Gomez M, Rodriguez-Manas L, Garcia FJ, Izquierdo M. Frailty assessment based on wavelet analysis during quiet standing balance test. J Biomech. 2011; 44:2213-20.

Mauritz KH, Dichgans J, Hufschmidt A. Quantitative analysis of stance in late cortical cerebellar atrophy of the anterior lobe and other forms of cerebellar ataxia. Brain. 1979; 102:461-82.

Mauritz KH, Dietz V. Characteristics of postural instability induced by ischemic blocking of leg afferents. Exp Brain Res. 1980; 38:117-9.

McCulloch KL, Buxton E, Hackney J, Lowers S. Balance, attention, and dual-task performance during walking after brain injury: associations with falls history. J Head Trauma Rehabil. 2010; 25:155-63.

McGimpsey G, Bradford TC. Limb prosthetics services and devices. Worcester, MA: Bioengineering Institute Center for Neuroprosthetics, Worcester Polytechnic Institution; 2010.

McIsaac TL, Lamberg EM, Muratori LM. Building a framework for a dual task taxonomy. Biomed Res Int. 2015; 2015:1-10.

McMulkin ML, Osebold WR, Mildes RD, Rosenquist RS. Comparison of three pediatric prosthetic feet during functional activities. J Prosthet Orthot. 2004; 16:78-84.

Menant JC, Schoene D, Sarofim M, Lord SR. Single and dual task tests of gait speed are equivalent in the prediction of falls in older people: a systematic review and meta-analysis. Ageing Res Rev. 2014; 16:83-104.

Metzger FG, Ehlis AC, Haeussinger FB, Schneeweiss P, Hudak J, Fallgatter AJ, et al. Functional brain imaging of walking while talking - An fNIRS study. Neuroscience. 2017; 343:85-93.

Miller LA, McCay JA. Summary and conclusion from the academy's sixth state-of-the-science conference on lower limb prosthetic outcome measures. SSC Proceedings. 2006; P2-P7.

Miller WC, Speechley M, Deathe AB. The prevalence and risk factors of falling and fear of falling among lower extremity amputees. Arch Phys Med Rehabil. 2001; 82:1031-7.

Misiti M, Misiti Y, Oppenheim G, Poggi JM. Wavelet Toolbox for use with Matlab. Wavelet Toolbox User's Guide: The MathWorks, Inc.; 1996.

Mohieldin A, Chidambaram A, Sabapathivinayagam R, Al Busairi W. Quantitative assessment of postural stability and balance between persons with lower limb amputation and normal subjects by using dynamic posturography. Maced J Med Sci. 2010; 3:138-43.

Montgomery DC, Peck EA, Vining GG. Introduction to linear regression analysis: John Wiley & Sons; 2012.

Morasso PG, Schieppati M. Can muscle stiffness alone stabilize upright standing? J Neurophysiol. 1999; 82:1622-6.

Morasso PG, Sanguineti V. Ankle muscle stiffness alone cannot stabilize balance during quiet standing. J Neurophysiol. 2002; 88:2157-62.

Morgan SJ, Hafner BJ, Kelly VE. The effects of a concurrent task on walking in persons with transfemoral amputation compared to persons without limb loss. Prosthet Orthot Int. 2016; 40:490-6.

Morgan SJ, Hafner BJ, Kelly VE. Dual-task walking over a compliant foam surface: A comparison of people with transfemoral amputation and controls. Gait Posture. 2017; 58:41-5.

Morris M, Iansek R, Matyas T, Summers J. Abnormalities in stride length-cadence relation in parkinsonian gait. Mov Disord. 1998; 13:61-9.

Muir-Hunter SW, Wittwer JE. Dual-task testing to predict falls in community-dwelling older adults: a systematic review. Physiotherapy. 2016; 102:29-40.

Muir SW, Speechley M, Wells J, Borrie M, Gopaul K, Montero-Odasso M. Gait assessment in mild cognitive impairment and Alzheimer's disease: The effect of dual-task challenges across the cognitive spectrum. Gait Posture. 2012; 35:96-100.

Murray MP, Kory RC, Clarkson BH, Sepic SB. Comparison of free and fast speed walking patterns of normal men. PMR. 1966; 45:8-24.

NLLIC. Diabetes and lower extremity amputations. In: America ACo, editor. http://www.amputee-coalition.org/fact_sheets/diabetes_leamp.html: National Limb Loss Information Center; 2008.

Nolan L, Wit A, Dudzinski K, Lees A, Lake M, Wychowanski M. Adjustments in gait symmetry with walking speed in trans-femoral and trans-tibial amputees. Gait Posture. 2003; 17:142-51.

Nordin E, Moe-Nilssen R, Ramnemark A, Lundin-Olsson L. Changes in step-width during dual-task walking predicts falls. Gait Posture. 2010; 32:92-7.

Olivier I, Cuisinier R, Vaugoyeau M, Nougier V, Assaiante C. Age-related differences in cognitive and postural dual-task performance. Gait Posture. 2010; 32:494-9.

Oppenheim U, Kohen-Raz R, Alex D, Kohen-Raz A, Azarya M. Postural characteristics of diabetic neuropathy. Diabetes care. 1999; 22:328-32.

Pajala S, Era P, Koskenvuo M, Kaprio J, Tormakangas T, Rantanen T. Force platform balance measures as predictors of indoor and outdoor falls in community-dwelling women aged 63-76 years. J Geront A. 2008; 63:171-8.

Parker K, Hanada E, Adderson J. Gait variability and regularity of people with transtibial amputations. Gait Posture. 2013; 37:269-73.

Pashler H. Graded capacity-sharing in dual-task interference? J Exp Psychol Hum Percept Perform. 1994; 20:330-42.

Pasquina PF, Cooper RA. Care of the Combat Amputee. Textbooks of Military Medicine. Washington, DC: Office of the Surgeon General; 2010. p. 820.

Patel P, Lamar M, Bhatt T. Effect of type of cognitive task and walking speed on cognitive-motor interference during dual-task walking. Neuroscience. 2014; 260:140-8.

Patla AE, Ishac MG, Winter DA. Anticipatory control of center of mass and joint stability during voluntary arm movement from a standing posture: interplay between active and passive control. Exp Brain Res. 2002; 143:318-27.

Paysant J, Beyaert C, Datie AM, Martinet N, Andre JM. Influence of terrain on metabolic and temporal gait characteristics of unilateral transtibial amputees. J Rehabil Res Dev. 2006; 43:153-60.

Peterka R. Sensorimotor integration in human postural control. J Neurophys. 2002; 88:1097-118.

Peters M. Footedness: asymmetries in foot preference and skill and neuropsychological assessment of foot movement. Psychol Bull. 1988; 103:179-92.

Plummer-D’Amato P, Altmann LJP, Saracino D, Fox E, Behrman AL, Marsiske M. Interactions between cognitive tasks and gait after stroke: A dual task study. Gait Posture. 2008; 27:683-8.

Plummer P, Eskes G. Measuring treatment effects on dual-task performance: a framework for research and clinical practice. Front Hum Neurosci. 2015; 9:225.

Reissland J, Manzey D. Serial or overlapping processing in multitasking as individual preference: Effects of stimulus preview on task switching and concurrent dual-task performance. Acta Psychol 2016; 168:27-40.

Resnik L, Borgia M. Reliability of outcome measures for people with lower-limb amputations: distinguishing true change from statistical error. Phys Ther. 2011; 91:555-65.

Roeing KL, Wajda DA, Sosnoff JJ. Time dependent structure of postural sway in individuals with multiple sclerosis. Gait Posture. 2016; 48:19-23.

Rosso AL, Cenciarini M, Sparto PJ, Loughlin PJ, Furman JM, Huppert TJ. Neuroimaging of an attention demanding dual-task during dynamic postural control. Gait Posture. 2017; 57:193-8.

Rusaw D, Hagberg K, Nolan L, Ramstrand N. Can vibratory feedback be used to improve postural stability in persons with transtibial limb loss? J Rehabil Res Dev. 2012; 49:1239-54.

Sadeghi H, Allard P, Prince F, Labelle H. Symmetry and limb dominance in able-bodied gait: a review. Gait Posture. 2000; 12:34-45.

Salavati M, Mazaheri M, Negahban H, Ebrahimi I, Jafari AH, Kazemnejad A, et al. Effect of dual-tasking on postural control in subjects with nonspecific low back pain. Spine. 2009; 34:1415-21.

Sample RB, Jackson K, Kinney AL, Diestelkamp WS, Reinert SS, Bigelow KE. Manual and Cognitive Dual Tasks Contribute to Fall-Risk Differentiation in Posturography Measures. J Appl Biomech. 2016; 32:541-7.

Sawers A, Hahn ME, Kelly VE, Czerniecki J, Kartin D. Beyond componentry: How principles of motor learning can enhance locomotor rehabilitation of individuals with lower limb loss—A review. J Rehabil Res Dev. 2012; 49:1431-42.

Sawers A, Hafner BJ. Outcomes associated with the use of microprocessor-controlled prosthetic knees among individuals with unilateral transfemoral limb loss: A systematic review. JRRD. 2013; 50:273-314.

Sawers AB, Hafner BJ. Outcomes associated with the use of microprocessor-controlled prosthetic knees among individuals with unilateral transfemoral limb loss: a systematic review. J Rehabil Res Dev. 2013; 50:273-314.

Schmalz T, Blumentritt S, Marx B. Biomechanical analysis of stair ambulation in lower limb amputees. Gait Posture. 2007; 25:267-78.

Schneiders AG, Sullivan J, O'Malley KJ, Clarke SV, Knappstein SA, Taylor LJ. A valid and reliable clinical determination of footedness. PMR. 2010; 2:835-41.

Schrager MA, Kelly VE, Price R, Ferrucci L, Shumway-Cook A. The effects of age on medio-lateral stability during normal and narrow base walking. Gait Posture. 2008; 28:466-71.

Scott DE, Dzendolet E. Quantification of sway in standing humans. Agressologie. 1972; 13:Suppl B:35-4.

Segal AD, Orendurff MS, Czerniecki JM, Shofer JB, Klute GK. Transtibial amputee joint rotation moments during straight-line walking and a common turning task with and without a torsion adapter. J Rehabil Res Dev. 2009; 46:375-83.

Segal AD, Orendurff MS, Czerniecki JM, Shofer JB, Klute GK. Local dynamic stability of amputees wearing a torsion adapter compared to a rigid adapter during straight-line and turning gait. J Biomech. 2010; 43:2798-803.

Sekiya N, Nagasaki H. Reproducibility of the walking patterns of normal young adults: test-retest reliability of the walk ratio(step-length/step-rate). Gait Posture. 1998; 7:225-7.

Seltman HJ. Experimental design and analysis. Pittsburgh: Carnegie Mellon University; 2012.

Shumway-Cook A, Woollacott M, Kerns KA, Baldwin M. The effects of two types of cognitive tasks on postural stability in older adults with and without a history of falls. J Geront A. 1997; 52:232-40.

Shurr D. Clinical perspectives on the prescription of prosthetic foot-ankle mechanisms. J Prosthet Orthot. 2005; 17:S31-S2.

Singh NK, Snoussi H, Hewson D, Duchêne J. Wavelet transform analysis of the power spectrum of centre of pressure signals to detect the critical point interval of postural control. Biomedical Engineering Systems and Technologies: Springer; 2010. p. 235-44.

Singh R, Hunter J, Philip A, Tyson S. Gender differences in amputation outcome. Disabil Rehabil. 2008; 30:122-5.

Sinha R, Van Den Heuvel WJ. A systematic literature review of quality of life in lower limb amputees. Disabil Rehabil. 2011; 33:883-99.

Smith DG, Michael JW, Bowker JH. Atlas of Amputations and Limb Deficiencies: Surgical, Prosthetic, and Rehabilitation Principles. 3rd ed: American Academy of Orthopaedic Surgeons; 2004.

Soames RW, Atha J. The spectral characteristics of postural sway behaviour. Eur J Appl Physiol Occup Physiol. 1982; 49:169-77.

Springer S, Giladi N, Peretz C, Yogev G, Simon ES, Hausdorff JM. Dual-tasking effects on gait variability: the role of aging, falls, and executive function. Mov Disord. 2006; 21:950-7.

Stark G. Perspectives on how and why feet are presecribed. J Prosthet Orthot. 2005; 17:S18-S22.

Swanenburg J, de Bruin ED, Favero K, Uebelhart D, Mulder T. The reliability of postural balance measures in single and dual tasking in elderly fallers and non-fallers. BMC Musculoskelet Disord. 2008; 9:162.

Taylor MJ, Strike SC, Dabnichki P. Turning bias and lateral dominance in a sample of able-bodied and amputee participants. Laterality. 2006; 12:50-63.

Thurner S, Mittermaier C, Hanel R, Ehrenberger K. Scaling-violation phenomena and fractality in the human posture control systems. Phys Rev E Stat Phys Plasmas Fluids Relat Interdiscip Topics. 2000; 62:4018-24.

Tombu M, Jolicoeur P. A central capacity sharing model of dual-task performance. J Exp Psychol Hum Percept Perform. 2003; 29:3-18.

Torrence C, Compo GP. A practical guide to wavelet analysis. Bull Amer Meteor. 1998; 79:61-78.

U. Raschke S, S. Orendurff M, Mattie JL, Kenyon DEA, Jones OY, Moe D, et al. Biomechanical characteristics, patient preference and activity level with different prosthetic feet: A randomized double blind trial with laboratory and community testing. J Biomech. 2015; 48:146-52.

van der Linde H, Hofstad CJ, Geurts AC, Postema K, Geertzen JH, van Limbeek J. A systematic literature review of the effect of different prosthetic components on human functioning with a lower-limb prosthesis. J Rehabil Res Dev. 2004; 41:555-70.

Vanicek N, Strike S, McNaughton L, Polman R. Gait patterns in transtibial amputee fallers vs. non-fallers: Biomechanical differences during level walking. Gait Posture. 2009; 29:415-20.

Vanicek N, Strike S, McNaughton L, Polman R. Postural responses to dynamic perturbations in amputee fallers versus nonfallers: a comparative study with able-bodied subjects. Arch Phys Med Rehabil. 2009; 90:1018-25.

Vrieling AH, van Keeken HG, Schoppen T, Otten E, Hof AL, Halbertsma JP, et al. Balance control on a moving platform in unilateral lower limb amputees. Gait Posture. 2008; 28:222-8.

Williams RM, Turner AP, Orendurff M, Segal AD, Klute GK, Pecoraro J, et al. Does having a computerized prosthetic knee influence cognitive performance during amputee walking? Arch Phys Med Rehabil. 2006; 87:989-94.

Winter DA. Human balance and posture control during standing and walking. Gait Posture. 1995; 3:193-214.

Winter DA, Patla AE, Ishac M, Gage WH. Motor mechanisms of balance during quiet standing. J Electromyogr Kinesiol. 2003; 13:49-56.

Woollacott M, Shumway-Cook A. Attention and the control of posture and gait: a review of an emerging area of research. Gait Posture. 2002; 16:1-14.

Wrightson JG, Twomey R, Ross EZ, Smeeton NJ. The effect of transcranial direct current stimulation on task processing and prioritisation during dual-task gait. Exp Brain Res. 2015; 233:1575-83.

Wrightson JG, Ross EZ, Smeeton NJ. The Effect of Cognitive-Task Type and Walking Speed on Dual-Task Gait in Healthy Adults. Motor Control. 2016; 20:109-21.

Yang YR, Chen YC, Lee CS, Cheng SJ, Wang RY. Dual-task-related gait changes in individuals with stroke. Gait Posture. 2007; 25:185-90.

Yazgan MY, Leckman JF, Wexler BE. A direct observational measure of whole body turning bias. Cortex. 1996; 32:173-6.

Yeung LF, Leung AK, Zhang M, Lee WC. Long-distance walking effects on trans-tibial amputees compensatory gait patterns and implications on prosthetic designs and training. Gait Posture. 2012; 35:328-33.

Yogev-Seligmann G, Hausdorff JM, Giladi N. The role of executive function and attention in gait. Mov Disord. 2008; 23:329-42.

Yogev-Seligmann G, Rotem-Galili Y, Mirelman A, Dickstein R, Giladi N, Hausdorff JM. How does explicit prioritization alter walking during dual-task performance? Effects of age and sex on gait speed and variability. Phys Ther. 2010; 90:177-86.

Yogev-Seligmann G, Hausdorff JM, Giladi N. Do we always prioritize balance when walking? Towards an integrated model of task prioritization. Mov Disord. 2012; 27:765-70.

Yogev-Seligmann G, Rotem-Galili Y, Dickstein R, Giladi N, Hausdorff JM. Effects of explicit prioritization on dual task walking in patients with Parkinson's disease. Gait Posture. 2012; 35:641-6.

Yogev G, Giladi N, Peretz C, Springer S, Simon ES, Hausdorff JM. Dual tasking, gait rhythmicity, and Parkinson's disease: which aspects of gait are attention demanding? Eur J Neurosci. 2005; 22:1248-56.

Ziegler-Graham K, MacKenzie EJ, Ephraim PL, Travison TG, Brookmeyer R. Estimating the prevalence of limb loss in the United States: 2005 to 2050. Arch Phys Med Rehabil. 2008; 89:422-9.

Zijlstra W, Rutgers AWF, Hof AL, Van Weerden TW. Voluntary and involuntary adaptation of walking to temporal and spatial constraints. Gait Posture. 1995; 3:13-8.

APPENDIX A

SUPPLEMENTAL ANALYSIS FOR CHAPTER 5

These results include the two intermediary standing conditions not included in the primary analysis. The results of this supplementary analysis further substantiate the results and conclusions reported in Chapter 5.

Supplement Table I

Mean (SD) for each sway parameter in single-task standing, no-prioritization dual-task standing, and prioritization dual-task standing for prosthesis users (PU) and controls (Ctrl) under different conditions (HS, hard surface; SS, soft surface; EO, eyes open; EC, eyes closed). Note HS/EO and SS/EC represent the Usual and Difficult standing conditions, respectively, in the main text. See Tables II and III for the results of statistical analysis.

Parameter/	Single-Task				No-Prioritization Dual-Task				Prioritization Dual-Task			
Group	HS/EO	HS/EC	SS/EO	SS/EC	HS/EO	HS/EC	SS/EO	SS/EC	HS/EO	HS/EC	SS/EO	SS/EC
PL												
PU	31.6 (12.9)	47.5 (25.2)	57.1 (20.9)	117.4 (61.9)	60.1 (33.5)	73.2 (43.2)	80.6 (42.8)	138.4 (78.5)	62.3 (58.6)	67.3 (46.3)	81.1 (48.5)	113.2 (51.6)
Ctrl	26.0 (9.0)	37.3 (17.2)	42.5 (15.9)	90.4 (56.0)	42.4 (13.3)	44.2 (13.0)	55.0 (15.3)	81.4 (37.7)	35.4 (11.9)	46.0 (16.0)	49.6 (12.3)	73.2 (30.6)
AREA												
PU	2.20 (0.90)	3.43 (3.93)	7.52 (5.04)	16.6 (12.6)	6.86 (5.87)	10.8 (16.0)	17.2 (14.9)	22.3 (14.4)	12.6 (18.9)	8.73 (10.7)	16.4 (15.2)	16.2 (9.0)
Ctrl	1.67 (1.09)	1.83 (1.57)	4.94 (3.49)	9.37 (5.79)	4.80 (3.64)	3.03 (2.09)	7.91 (5.15)	8.54 (5.46)	4.30 (5.89)	4.93 (6.61)	6.82 (4.38)	7.35 (3.60)
AP												
PU	1.1 (0.5)	1.3 (0.8)	2.4 (1.1)	3.7 (1.5)	2.9 (2.1)	2.9 (2.5)	4.7 (3.0)	4.7 (2.1)	3.3 (3.3)	3.2 (2.9)	4.5 (3.1)	4.0 (2.0)
Ctrl	0.93 (0.4)	0.88 (0.3)	1.6 (0.7)	2.2 (0.73)	1.8 (0.9)	1.3 (0.6)	2.4 (1.0)	2.2 (1.0)	1.8 (1.5)	1.6 (1.1)	2.3 (1.5)	2.0 (0.74)
ML												
PU	2.4 (0.6)	3.0 (1.1)	4.1 (0.74)	6.0 (1.1)	3.7 (1.7)	4.0 (1.5)	4.5 (1.4)	6.5 (2.1)	3.8 (2.6)	3.5 (1.2)	5.1 (1.6)	6.0 (6.0)
Ctrl	2.5 (1.0)	2.7 (1.0)	6.0 (1.1)	5.6 (1.5)	3.6 (1.8)	3.0 (0.8)	4.0 (1.2)	4.9 (1.4)	2.8 (1.3)	3.2 (1.6)	4.0 (1.0)	4.8 (1.5)

PL, path length (cm); AREA, 95% area (cm^2); AP, anterior-posterior amplitude (cm); ML, medial-lateral amplitude (cm)

Supplement Table II. Summary of p-values for the main effects and 2-way interactions from the 3-way ANOVA in the single-task standing condition for each sway parameter (3-way interactions not shown because of no significance, p= 0.141 – 0.597). Significant values in bold. The results indicate increased sway with increasing postural challenge.

Parameter	Main Effect			Interaction		
	Group	Surf	Vis	Group-Surf	Group-Vis	Surf-Vis
PL	0.153	**0.001**	**0.001**	0.184	0.405	**0.001**
AREA	0.066	**0.001**	**0.001**	0.055	0.081	**0.001**
AP	**0.006**	**0.001**	**0.001**	**0.009**	**0.020**	**0.001**
ML	0.319	**0.001**	**0.001**	0.112	0.977	**0.001**

PL, path length (cm); AREA, 95% area (cm^2); AP, anterior-posterior amplitude (cm); ML, medial-lateral amplitude (cm); Surf, surface; Vis, vision.

Supplement Table III. Summary of p-values for the main effects and 2-way interactions from the within-group 3-way ANOVA between the single-task and no-prioritization dual-task standing and the different standing conditions for each sway parameter in prosthesis users (PU) and controls (Ctrl), respectively (3-way interactions not shown because of no significance, p= 0.219 – 0.978). Significant values in bold. The results support a decrease in sway for controls and an increase for prosthesis users.

Parameter/		Main Effect			Interaction		
Group		Task	Surf	Vis	Task-Surf	Task-Vis	Surf-Vis
PL							
	PU	**0.002**	**0.001**	**0.001**	0.586	0.631	**0.001**
	Ctrl	0.068	**0.001**	**0.001**	**0.046**	**0.014**	**0.001**
AREA							
	PU	**0.003**	**0.001**	**0.033**	0.625	0.835	0.225
	Ctrl	**0.016**	**0.001**	0.078	0.274	**0.008**	**0.001**
AP							
	PU	**0.003**	**0.001**	0.185	0.964	0.074	0.389
	Ctrl	**0.002**	**0.001**	0.808	0.226	**0.016**	**0.049**
ML							
	PU	**0.012**	**0.001**	**0.001**	0.211	0.762	**0.002**
	Ctrl	0.070	**0.001**	**0.003**	0.077	**0.002**	**0.001**

PL, path length (cm); AREA, 95% area (cm^2); AP, anterior-posterior amplitude (cm); ML, medial-lateral amplitude (cm); Surf, surface; Vis, vision.

Supplement Table IV. Means (SDs) of dual-task cost for each sway parameter in prosthesis users (PU) and controls (Ctrl) without (no-prioritization) and with (prioritization) instruction to focus on improving cognitive task performance under different standing conditions (HS, hard surface; SS, soft surface; EO, eyes open; EC, eyes closed). See Table V for the results of statistical analysis.

Parameter/	No-Prioritization Dual-Task Cost				Prioritization Dual-Task Cost			
Group	HS/EO	HS/EC	SS/EO	SS/EC	HS/EO	HS/EC	SS/EO	SS/EC
PL								
PU	-28.5 (28.9)	-25.6 (30.2)	-23.4 (28.7)	-20.9 (32.8)	-30.7 (50.7)	-19.7 (36.0)	-23.9 (31.8)	4.3 (22.7)
Ctrl	-16.3 (15.7)	-6.9 (14.5)	-12.5 (10.0)	9.0 (35.3)	-9.3 (15.7)	-8.7 (18.6)	-7.1 (10.2)	17.2 (37.4)
AREA								
PU	-4.67 (5.61)	-7.40 (16.2)	-9.68 (13.8)	-5.72 (7.77)	-10.44 (18.4)	-5.31 (9.98)	-8.92 (12.3)	0.38 (6.60)
Ctrl	-3.13 (3.79)	-1.20 (2.38)	-2.97 (3.78)	0.83 (5.08)	-2.63 (5.64)	-3.09 (5.23)	-1.89 (3.91)	2.01 (5.01)
AP								
PU	-1.8 (2.1)	-1.6 (2.6)	-2.4 (2.9)	-0.94 (0.8)	-2.2 (3.2)	-1.8 (3.1)	-2.2 (2.6)	-0.30 (1.4)
Ctrl	-0.84 (0.7)	-0.44 (0.7)	-0.76 (1.1)	0.05 (0.9)	-0.86 (1.5)	-0.77 (0.9)	-0.65 (1.6)	0.23 (0.8)
ML								
PU	-1.2 (1.6)	-1.0 (1.8)	-0.41 (1.0)	-0.48 (1.5)	-1.4 (2.3)	-0.48 (1.3)	-1.0 (1.3)	-0.03 (1.1)
Ctrl	-1.1 (1.9)	-0.31 (1.3)	-0.61 (1.0)	0.73 (1.0)	-0.35 (1.1)	-0.53 (1.1)	-0.59 (0.8)	0.85 (1.3)

PL, path length (cm); AREA, 95% area (cm^2); AP, anterior-posterior amplitude (cm); ML, medial-lateral amplitude (cm)

Supplement Table V. Summary of p-values for the main effects and 2-way interactions from the within-group 3-way ANOVA on dual-task cost for each sway parameter in prosthesis users (PU) and controls (Ctrl), respectively (3-way interactions not shown because of no significance, p= 0.079 – 0.789). Significant values in bold. The results show no change or reduced sway in the prioritization condition.

Parameter/ Group		Main Effect			Interaction		
		Instruction	Surf	Vis	Instruction-Surf	Instruction-Vis	Surf-Vis
PL							
	PU	0.109	0.235	0.058	0.292	0.148	0.407
	Ctrl	**0.035**	**0.013**	**0.021**	0.294	0.240	0.072
AREA							
	PU	0.690	0.719	0.154	0.149	0.124	0.198
	Ctrl	0.708	0.085	**0.007**	0.268	0.216	0.130
AP							
	PU	0.863	0.499	**0.018**	0.093	0.678	0.138
	Ctrl	0.897	0.135	**0.011**	0.347	0.608	0.125
ML							
	PU	0.747	0.206	0.061	0.637	**0.047**	0.811
	Ctrl	0.405	**0.040**	**0.002**	0.619	0.089	0.079

PL, path length (cm); AREA, 95% area (cm^2); AP, anterior-posterior amplitude (cm); ML, medial-lateral amplitude (cm); Surf, surface; Vis, vision.

APPENDIX B

FULL TABLES FOR CHAPTER 6

Table I. Percent contribution to total spectral power (%) from different frequency bands (mean and SE) in the ML direction during single-task and dual-task standing. The standing condition and the cognitive task both impacted the percent contribution from each frequency band. There was no difference between groups in percent contribution.

Standing Condition	Task	Very Low [0 – 0.19 Hz]		Low [0.19 – 0.39 Hz]		Middle [0.39 – 1.17 Hz]		High [1.17 – 10.15 Hz]	
		Prosthesis	Control	Prosthesis	Control	Prosthesis	Control	Prosthesis	Control
Eyes Open/Hard Surface									
	Single	75 (3)	75 (3)	12 (2)	12 (2)	8 (1)	9 (1)	4 (1)	4 (1)
	Dual	59 (7)	59 (6)	12 (3)	17 (4)	17 (3)	16 (3)	11 (3)	8 (2)
Eyes Closed/Hard Surface									
	Single	70 (3)	68 (5)	11 (1)	10 (1)	11 (1)	13 (2)	7 (1)	8 (2)
	Dual	63 (4)	59 (6)	9 (2)	15 (3)	17 (2)	16 (3)	11 (2)	9 (2)
Eyes Open/Soft Surface									
	Single	67 (4)	67 (3)	18 (3)	16 (2)	11 (1)	11 (1)	4 (1)	5 (1)
	Dual	60 (5)	56 (4)	24 (3)	24 (2)	11 (2)	14 (3)	5 (1)	5 (1)
Eyes Closed/Soft Surface									
	Single	58 (4)	57 (5)	16 (2)	16 (2)	17 (2)	17 (2)	8 (1)	10 (3)
	Dual	49 (5)	52 (6)	19 (2)	20 (3)	20 (3)	19 (3)	12 (1)	9 (2)
ANOVA Single-Task									
Main Effects	Vision	**0.005**		0.417		**<0.001**		**<0.001**	
	Surface	**<0.001**		**0.003**		**0.001**		0.080	
	Group	0.876		0.614		0.702		0.580	
2-way Interactions	Vison x Group	0.718		0.712		0.855		0.774	
	Surface x Group	0.844		0.970		0.592		0.942	
	Vision x Surface	0.204		0.597		0.152		0.651	
3-way Interaction	Vision x Surface x Group	0.967		0.533		0.495		0.873	
ANOVA Dual-Task									
Main Effects	Task	**<0.001**		**0.004**		**0.001**		**0.009**	
2-way Interactions	Task x Vision	0.238		0.298		0.463		0.416	
	Task x Surface	0.255		0.130		0.034		0.047	
	Task x Group	0.860		0.086		0.835		0.110	
3-way Interactions	Task x Vision x Group	0.643		0.955		0.453		0.726	
	Task x Surface x Group	0.926		0.234		0.519		0.579	
	Task x Vision x Surface	0.406		0.470		0.187		0.031	
4-way Interaction	Task x Vision x Surface x Group	0.476		0.801		0.699		0.106	

Table II. The relative spectral power (%) from different frequency bands (mean and SE) in the AP direction during single-task and dual-task standing. The standing condition impacted the relative contribution from each frequency band, however, performing the cognitive task did not. There was no difference between groups in relative contribution.

Standing Condition	Task	Very Low [0 – 0.19 Hz]		Low [0.19 – 0.39 Hz]		Middle [0.39 – 1.17 Hz]		High [1.17 – 10.15 Hz]	
		Prosthesis	Control	Prosthesis	Control	Prosthesis	Control	Prosthesis	Control
Eyes Open/Hard Surface									
	Single	70 (3)	76 (3)	9 (1)	8 (2)	14 (2)	11 (1)	6 (1)	5 (1)
	Dual	67 (7)	75 (5)	10 (2)	12 (4)	17 (4)	9 (2)	6 (2)	5 (1)
Eyes Closed/Hard Surface									
	Single	62 (5)	70 (4)	8 (2)	9 (2)	18 (3)	13 (2)	12 (3)	8 (1)
	Dual	65 (6)	69 (6)	10 (2)	7 (2)	15 (4)	13 (3)	8 (2)	10 (4)
Eyes Open/Soft Surface									
	Single	63 (5)	66 (5)	13 (3)	17 (3)	17 (3)	13 (3)	7 (2)	3 (1)
	Dual	59 (7)	65 (5)	18 (3)	17 (2)	16 (3)	13 (3)	6 (2)	3 (1)
Eyes Closed/Soft Surface									
	Single	46 (5)	56 (6)	18 (3)	17 (2)	24 (2)	20 (4)	12 (2)	7 (1)
	Dual	62 (5)	60 (6)	14 (3)	17 (3)	15 (3)	18 (4)	8 (2)	5 (1)
ANOVA Single-Task									
Main Effects	Vision	**<0.001**		0.278		**0.001**		**<0.001**	
	Surface	**0.003**		**0.001**		**0.030**		0.813	
	Group	0.107		0.780		0.091		0.032	
2-way Interactions	Vison x Group	0.374		0.478		0.667		0.306	
	Surface x Group	0.884		0.575		0.961		0.627	
	Vision x Surface	0.379		0.365		0.350		0.845	
3-way Interaction	Vision x Surface x Group	0.751		0.196		0.827		0.768	
ANOVA Dual-Task									
Main Effects	Task	0.597		0.491		0.307		0.410	
2-way Interactions	Task x Vision	0.056		0.147		0.094		0.427	
	Task x Surface	0.382		0.701		0.415		0.420	
	Task x Group	0.641		0.951		0.741		0.347	
3-way Interactions	Task x Vision x Group	0.192		0.876		0.103		0.257	
	Task x Surface x Group	0.808		0.922		0.406		0.597	
	Task x Vision x Surface	0.411		0.516		0.454		0.478	
4-way Interaction	Task x Vision x Surface x Group	0.664		0.072		0.805		0.510	

Table III. The relative spectral power (%) from different frequency bands (mean and SE) recorded on the prosthetic and intact sides in the ML direction during single-task and dual-task standing. The prosthetic side had lower contribution from the middle and high frequency bands and higher contribution from the very low frequency band. Only the high frequency band had a significant Side interactions, showing less asymmetry while dual-tasking with eyes closed.

Standing Condition	Task	Very Low [0 – 0.19 Hz)		Low [0.19 – 0.39 Hz)		Middle [0.39 – 1.17 Hz)		High [1.17 – 10.15 Hz)	
		Prosthetic	Intact	Prosthetic	Intact	Prosthetic	Intact	Prosthetic	Intact
Eyes Open/Hard Surface									
	Single	81 (3)	72 (3)	12 (2)	13 (2)	6 (1)	10 (1)	1 (0.2)	5 (1)
	Dual	68 (7)	53 (7)	11 (3)	14 (3)	12 (3)	19 (3)	9 (3)	14 (3)
Eyes Closed/Hard Surface									
	Single	79 (2)	67 (3)	12 (2)	10 (1)	6 (1)	13 (1)	3 (0.4)	9 (1)
	Dual	66 (5)	62 (5)	9 (2)	9 (2)	15 (3)	18 (2)	10 (3)	11 (2)
Eyes Open/Soft Surface									
	Single	73 (4)	64 (4)	17 (3)	18 (3)	7 (1)	12 (1)	2 (0.3)	6 (1)
	Dual	69 (5)	58 (5)	19 (4)	24 (4)	9 (2)	11 (2)	4 (1)	6 (1)
Eyes Closed/Soft Surface									
	Single	67 (4)	55 (4)	16 (2)	16 (2)	12 (2)	19 (2)	4 (1)	10 (1)
	Dual	58 (5)	48 (6)	18 (3)	18 (3)	16 (2)	21 (3)	7 (1)	13 (1)
ANOVA									
Main Effects	Side	**<0.001**		0.228		**<0.001**		**<0.001**	
2-way Interactions	Side x Task	0.849		0.154		0.330		0.394	
	Side x Vision	0.706		0.110		0.604		**0.010**	
	Side x Surface	0.837		0.599		0.705		0.861	
3-way Interactions	Side x Task x Vision	0.071		0.309		0.111		**0.012**	
	Side x Task x Surface	0.935		0.764		0.664		0.710	
	Side x Vision x Surface	0.544		0.644		0.388		0.233	
4-way Interaction	Side x Task x Vision x Surface	0.417		0.403		0.241		0.083	

Table IV. The relative spectral power (%) from different frequency bands (mean and SE) recorded on the prosthetic and intact sides in the AP direction during single-task and dual-task standing. Only the high frequency band had higher percent contribution on the intact side compared to the prosthetic side.

Standing Condition	Task	Very Low [0 – 0.19 Hz)		Low [0.19 – 0.39 Hz)		Middle [0.39 – 1.17 Hz)		High [1.17 – 10.15 Hz)	
		Prosthetic	Intact	Prosthetic	Intact	Prosthetic	Intact	Prosthetic	Intact
Eyes Open/Hard Surface									
	Single	80 (2)	76 (4)	9 (2)	10 (2)	8 (1)	9 (1)	2 (0.5)	5 (1)
	Dual	67 (6)	60 (8)	12 (2)	13 (3)	14 (4)	17 (3)	7 (2)	10 (3)
Eyes Closed/Hard Surface									
	Single	76 (4)	69 (4)	11 (2)	11 (2)	9 (1)	11 (2)	4 (1)	7 (1)
	Dual	73 (6)	74 (4)	9 (2)	7 (2)	11 (3)	11 (2)	6 (2)	7 (2)
Eyes Open/Soft Surface									
	Single	69 (4)	66 (4)	17 (3)	15 (3)	9 (1)	12 (1)	4 (1)	6 (1)
	Dual	68 (5)	62 (5)	20 (3)	20 (4)	8 (1)	11 (1)	4 (1)	6 (1)
Eyes Closed/Soft Surface									
	Single	67 (5)	55 (4)	17 (3)	16 (3)	11 (2)	18 (2)	5 (1)	11 (1)
	Dual	68 (6)	50 (6)	13 (2)	17 (3)	13 (4)	18 (3)	6 (2)	14 (2)
ANOVA									
Main Effects	Side	0.112		0.769		0.103		**0.022**	
2-way Interactions	Side x Task	0.830		0.511		0.827		0.990	
	Side x Vision	0.386		0.897		0.484		0.054	
	Side x Surface	0.151		0.897		0.072		0.144	
3-way Interactions	Side x Task x Vision	0.716		0.865		0.068		0.908	
	Side x Task x Surface	0.169		0.097		0.815		0.543	
	Side x Vision x Surface	0.055		0.407		0.249		0.031	
4-way Interaction	Side x Task x Vision x Surface	0.514		0.638		0.626		0.302	

APPENDIX C

RESIDUAL STANDARD DEVIATION MATHMATICAL DERIVATION

Five descriptor variables of an XY dataset (means standard deviations, correlation) can be used to derive the RSD calculation. The variable of interest (ex. stride length or cadence) is designated as Y. The correlated variable (ex. velocity) is designated as X.

$$M_y = Mean\ Y$$

$$M_x = Mean\ X$$

$$\sigma_y = Sample\ Standard\ deivation\ of\ Y$$

$$\sigma_x = Sample\ Standard\ deivation\ of\ X$$

$$r = Pearson's\ r\ correlation\ between\ X\ and\ Y$$

Step 1: Calculate the best fit line of the XY data set to identify the slope and intercept. This line provides a set of predicted Y values (f) for every X value.

$$f_i = bX_i + A$$

$$b = slope$$

$$A = intercept$$

The slope and intercept of the best fit line are calculated from the 5 descriptor variables.

$$b = r({}^{\sigma_y}/_{\sigma_x})$$

$$A = M_y - bM_x$$

Or

$$A = M_y - r({}^{\sigma_y}/_{\sigma_x})M_x$$

For simplification purposes the intercept A term can be represented as A in the following derivation. Thus, substituting b into the linear equation gives:

$$f_i = r\left({}^{\sigma_y}/_{\sigma_x}\right)X_i + A$$

Step 2: Calculate the residuals, R_i:

$$R_i = Y_i - f_i$$

$$R_i = Y_i - (r\left({}^{\sigma_y}/_{\sigma_x}\right)X_i + A)$$

Step 3: Find the standard deviation of the residuals, *RSD*:

$$RSD = \sqrt{\frac{1}{N}\sum_{i=1}^{N}(R_i - \bar{R})^2}$$

where

$$\bar{R} = \frac{\sum R_i}{N}$$

$$\bar{R} = \frac{\sum Y_i - (r\left({}^{\sigma_y}/_{\sigma_x}\right)X_i + A)}{N}$$

Substituting the formulas for R_i and $\overline{R}$ into the standard deviation equation gives:

$$RSD = \sqrt{\frac{1}{N}\sum_{i=1}^{N}[Y_i - (r\left({}^{\sigma_y}/_{\sigma_x}\right)X_i + A) - \frac{\sum(Y_i - (r\left({}^{\sigma_y}/_{\sigma_x}\right)X_i + A)}{N}]^2}$$

The equation can be simplified to:

$$RSD = \sqrt{\frac{1}{N}\sum_{i=1}^{N}[Y_i - r\left({}^{\sigma_y}/_{\sigma_x}\right)X_i - \frac{\sum Y_i - r({}^{\sigma_y}/_{\sigma_x})X_i}{N}]^2}$$

The intercept term A cancels out of the equation. The slope term, representing the weighted normalized variability, remains as the primary term in the RSD calculation. This is the baseline calculation. For the dual-task calculation, the coordinates for the dual-task walk are used for Y_i, X_i, and N. The r, σ_y, σ_x, M_y, and M_x values are used from the baseline data.

The strength of the linear relationship determines how much the RSD calculation differs from the traditional SD calculation. As r (the correlation between the X and Y variables) approaches zero the term from the best fit line equation will dissipate and the RSD calculation approaches the calculation for the standard deviation of Y.

$$RSD = \sqrt{\frac{1}{N}\sum_{i=1}^{N}[Y_i - 0 * \left({}^{\sigma_y}/_{\sigma_x}\right)X_i - \frac{\sum Y_i - 0 * ({}^{\sigma_y}/_{\sigma_x})X_i}{N}]^2}$$

$$RSD = \sqrt{\frac{1}{N}\sum_{i=1}^{N}(Y_i - \frac{\sum Y_i}{N})^2}$$

$$RSD = \sqrt{\frac{1}{N}\sum_{i=1}^{N}(Y_i - \bar{Y})^2} = SD\ of\ Y$$

Thus, RSD provides a measure of variability that accounts for the dependence of one variable on the other. It can be used for any XY data set, but it may be most useful when X and Y variables are highly correlated.

APPENDIX D

CO-AUTHOR PERMISSION

To whom it may concern,

The below signed give permission for the published manuscripts:

"Lower limb preference on goal-oriented tasks in unilateral prosthesis users"

"Stride length-cadence relationship is disrupted in below-knee prosthesis users"

"Residual standard deviation: validation of a new measure of dual-task cost in below-knee prosthesis users"

"Increased alertness, better than posture prioritization, explains dual-task performance in prosthesis users and controls under increasing postural and cognitive challenge"

which were written in conjunction with work conducted at Methodist Rehabilitation Center in Jackson, MS, to be included in Charla Lindley Howard's thesis dissertation. Each signee gives permission for use of all listed papers on which he/she was listed as a co-author.

10/4/2017
Dobrivoje Stokic

10/4/2017
Chris Wallace

10/4/2017
John Chow

10/4/2017
Bonnie Perry

Ira A. Fulton Schools of Engineering
Arizona State University

16 October 2017

To whom it may concern:

I give my permission for the following published manuscript to be included in Charla Lindley Howard's dissertation:

> "C. Howard, C. Wallace, J. Abbas, and D. Stokic, "Residual Standard Deviation: Validation of a New Measure of Dual-Task Cost in Below-Knee Prosthesis Users", *Gait & Posture*, Vol 15, pp. 91-96, (doi:dx.doi.org/10.1016/j.gaitpost.2016.09.025), 2017.

Please contact me if you would like further information.

Sincerely,

James J. Abbas, PhD
Associate Professor, School of Biological and Health Systems Engineering

School of Biological and Health Systems Engineering
P.O. Box 879709 Tempe, AZ 85287-9709
p: 480-965-3028 **f:** 480-727-7624 **web:** sbhse.engineering.asu.edu

APPENDIX E

IRB APPROVAL

Earl R. Wilson, Founding Chairman
METHODIST
REHABILITATION CENTER

October 26, 2010

Dobrivoje Stokic, M.D., D.Sc.
Administrative Director for Research
Methodist Rehabilitation Center
1350 East Woodrow Wilson
Jackson, MS 39216

Chris Wallace, CPO, FAAOP
Methodist Rehabilitation Center
One Layfair Drive, Suite 300
Flowood, MS 39232

RE: Movement Control Strategies during Performance of Balance and Action Tasks in Lower Limb Prosthesis Users *[MRC RP # 2010-05]*

Dear Dr. Stokic and Mr. Wallace:

Thank you for the submission of the above-named protocol to the Institutional Review Board (IRB) of the Methodist Rehabilitation Center (MRC). The investigational research study complies with the requirements and qualifications for expedited review in accordance with 45 CFR §46.110(b) and the Informed Consents meet the requirements of the IRB.

The protocol meets the approval of the IRB; therefore, the MRC IRB is pleased to inform you that the above named study is granted full approval.

Any changes to the existing research protocol or the Informed Consents must receive IRB approval prior to implementation. Approval for the study expires one year from the above date. The research investigator is responsible to assure that the study is not conducted beyond the expiration date without IRB review and approval for continuation. You may renew the study at that time upon completion of the MRC IRB Annual Review form. If you close the study prior to the expiration date, please inform the IRB chairperson.

If there are any questions, do not hesitate to contact me at (601) 364-3542.

Sincerely,

Tammy M. Voynik, J.D.
Vice President Legal Affairs
Chairperson, Institutional Review Board
(FWA-#00003957)

tmv

1350 E. Woodrow Wilson, Jackson, Mississippi 39216
Telephone : 601.981.2611 | Toll-free : 1.800.223.6672 | www.methodistonline.org

Earl R. Wilson, Founding Chairman
METHODIST
REHABILITATION CENTER

June 28, 2011

Dobrivoje Stokic, M.D., D.Sc.
Administrative Director for Research
Methodist Rehabilitation Center
1350 East Woodrow Wilson
Jackson, MS 39216

Chris Wallace, CPO, FAAOP
Methodist Rehabilitation Center
One Layfair Drive, Suite 300
Flowood, MS 39232

RE: "Talking While Walking": Dual Task Performance in Lower Limb Prosthesis Users *[MRC RP # 2011-01]*

Dear Dr. Stokic and Mr. Wallace:

Thank you for the submission of the above-named protocol to the Institutional Review Board (IRB) of the Methodist Rehabilitation Center (MRC). The investigational research study complies with the requirements and qualifications for expedited review in accordance with 45 CFR §46.110(b) and the Informed Consent meets the requirements of the IRB.

The protocol meets the approval of the IRB; therefore, the MRC IRB is pleased to inform you that the above named study and related Informed Consent is granted full approval. You must obtain IRB approval of any separate/additional Informed Consent form if additional screening or testing of diabetic participants as described in the study is required.

Any changes to the existing research protocol or the Informed Consent must receive IRB approval prior to implementation. Approval for the study expires one year from the above date. The research investigator is responsible to assure that the study is not conducted beyond the expiration date without IRB review and approval for continuation. You may renew the study at that time upon completion of the MRC IRB Annual Review form. If you close the study prior to the expiration date, please inform the IRB chairperson.

If there are any questions, do not hesitate to contact me at (601) 364-3542.

Sincerely,

Tammy M. Voynik, J.D.
Vice President Legal Affairs
Chairperson, Institutional Review Board
(FWA-#00003957)

tmv

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MMRC INSTITUTIONAL REVIEW BOARD
APPLICATION FOR EXPEDITED REVIEW

APPROVED
JUL 15 2013
MRC IRB 2013-02
MRC IRB

This form is to be completed and submitted to the IRB of the research protocol only when the investigator is contemplating the initiation of a research project which, in the investigator's judgment, qualifies for expedited review (If additional space is necessary, attach a separate sheet).

A. DATE OF APPLICATION:

July 9, 2013

B. PRINCIPAL INVESTIGATOR(s) and CO-PRINCIPAL INVESTIGATOR(s):

Co-Principal Investigators: Chris Wallace, CPO, FAAOP and Dobrivoje S. Stokic, MD, DSc

Co-Investigator: Charla Howard

C. COMPLETE MAILING ADDRESS AND PHONE NUMBER OF PI(s) and CO-PI(s):

D.S. Stokic and C. Howard
Methodist Rehabilitation Center
1350 East Woodrow Wilson
Jackson, MS 39216
601-364-3314

C. Wallace
Methodist Rehabilitation Center
One Layfair Dr., Suite 300
Flowood, Mississippi 39232
601-936-8899

D. TITLE OF PROJECT:

Dual-task performance of lower limb prosthesis users during standing and walking

E. EXPECTED STARTING DATE (No research may be initiated until certification is granted):

July 15, 2013

F. PREDICTED COMPLETION DATE (Include all aspects of research and final write-up):

March 1, 2014

G. ANTICIPATED COST TO INSTITUTION:

$130 for two Airex Balance Pads, reimbursement of $20 each for up to 50 prosthesis users and 25 control subjects who are not MRC employees. $6 lunch ticket at cafeteria for all subjects.

H. OBJECTIVES OF PROJECT (BRIEFLY STATE THE PURPOSE OF THE RESEARCH, WITH SPECIAL REFERENCE TO HUMAN SUBJECTS INVLOVED)

- Examine the effects of a concurrent cognitive task on postural sway and walking pattern (dual-task paradigm) in lower limb prosthesis users in comparison to healthy subjects, and
- Evaluate the influence of task prioritization by altering the difficulty of postural (eyes closed, soft surface, tandem stance) and gait (wide and narrow path) tasks with and without implicit instruction.

We hypothesize that greater postural challenge will shift resources toward maintenance of stability and away from performance of the concurrent task. We also predict that prosthesis users will show a greater demand on resources for posture and gait tasks than control subjects.

APPENDIX F

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Author of new book	Charla Lindley Howard
Expected publication date of new book	Dec 2017
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BIOGRAPHICAL SKETCH

Charla Lindley Howard graduated from Mississippi State University with a Bioengineering degree in 2008. While pursuing her graduate degree from Arizona State University, she worked as a research associate at Methodist Rehabilitation Hospital in Jackson, MS. She also earned her prosthetist certificate from the Newington Certificate Program in Orthotics and Prosthetics. Charla has a passion for prosthetic research and plans to continue her work to help improve access to prosthetic devices that allow lower-limb amputees to meet and exceed their mobility goals.