Transtibial amputee and able-bodied walking strategies for maintaining stable gait in a multi-terrain virtual reality environment

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Abstract

The CAREN-Extended system is a fully immersive virtual environment (VE) that can provide stability-challenging scenarios in a safe, controlled manner. Understanding gait biomechanics when stability is challenged is required when developing quantifiable metrics for rehabilitation assessment.

The first objective of this thesis was to examine the VE's technical aspects to ensure data validity and to design a stability-challenging VE scenario. The second and third objectives examined walking speed changes and kinematic strategies when stability was challenged for able-bodied and unilateral transtibial amputees.

The results from this thesis demonstrated: 1) understanding VE operating characteristics are important to ensure data validity and to effectively design virtual scenarios; 2) self-paced treadmill mode for VEs with multiple movement scenarios may elicit more natural gait; 3) gait variability and trunk motion measures are useful when quantitatively assessing stability performance for people with transtibial amputations.

Table of Contents

List of Figures

List of Tables

Abbreviations and definitions

AB	Able-bodied participants with no amputation or injury that affects walking ability
AP	Anterior-posterior
BOS	Base of support, defined by the feet on the support surface
BS	Bottom-cross-slope, right- and left-cross-slope gait cycles where the limb is at the bottom of the slope
COM	Centre of mass, 3D position of the body's centre of gravity
Contralateral	Referring to the opposite side of the body
D-Flow	D-Flow is the control software for the CAREN virtual environment
Dominant limb	Limb used to kick a ball (AB)
DS	7° downhill treadmill walking
DST	Double support time, when two feet are in contact with the support surface
FA_{FS}	Foot-surface angle at foot strike
FA_{TO}	Foot-surface angle at toe off
Foot strike	Foot contacts the support surface
Gait cycle	Walking cycle that includes swing and stance phases
HL	Rolling hills treadmill walking
Intact limb	Unaffected or sound limb (TT)
Ipsilateral	Referring to the same side of the body
Kinematics	Movement of joints or body segments
Kinetics	Forces and moments that produce movement
K-level 3	Amputee with the ability or potential for ambulation with variable cadence. Typical of the community ambulator who has the ability to traverse beyond simple locomotion.
K-level 4	Amputee has the ability or potential for prosthetic ambulation that exceeds basic ambulation skills, exhibiting high impact, stress, or energy levels.

LS	5° left-cross-slope level treadmill walking
LW	Level treadmill walking
MaxFC	Maximum foot clearance
MinFC	Minimum foot clearance
ML	Medial-lateral
ML-MoS	ML margin-of-stability
MLT	Level treadmill walking with continuous ML translations
mSD	Mean of the standard deviations
Non-dominant limb	Opposite limb to the foot used to kick a ball (AB)
Prosthetic limb	Amputated limb with a prosthesis (TT)
R	Statistical analysis software (The R Project for Statistical Computing)
RMS_T	Root mean square of trunk acceleration
RO	Simulated rocky treadmill walking
RS	5° right-cross-slope treadmill walking
SL	Step length
SP	Self-paced treadmill mode
ST	Step time
Stance	Phase with foot in contact with the support surface
Step	Foot strike to foot strike of the other limb
Stride	Foot strike to subsequent foot strike on the same limb
SW	Step width
Swing	Phase with foot not in contact with the support surface
Toe off	Foot leaves the support surface
TS	Top-cross-slope, right- and left-cross-slope gait cycles where the limb is at the top of the slope
TT	Unilateral transtibial amputee participants who have sustained an amputation below the knee on one limb
US	7° uphill treadmill walking
Visual3D	Biomechanics analysis software for analyzing data collected with a 3D motion capture system
VT	Vertical

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Chapter 1: Introduction

1.1 Introduction

The challenge of bipedal walking is to actively balance the body over the base of support (feet) while producing forward momentum and maintaining upright posture. Furthermore, one limb supports the body for a significant portion of the gait cycle and the opportunity to lose balance and fall during single limb support is greater. This dynamic balancing task requires a complex postural control system involving vestibular, visual, and somatosensory subsystems. If the subsystem(s) are adversely affect by lower limb amputation, knee osteoarthritis, lower limb neuropathy, brain injury, and aging [1-8], the ability to walk is also affected. Walking becomes more challenging and is linked with increased energy consumption [9, 10], decreased walking stability (which increases fall risk [11-15]), and compensatory gait strategies that may lead to future health problems. Thus, a good understanding of dynamic stability for people with mobility-related disabilities is important when making healthcare decisions.

With advancements in virtual environments (VE), the biomechanics of human walking stability can be examined in a safe and controlled, environment that approximates real-world conditions. The Computer Aided Rehabilitation Environment (CAREN, Motek Medical, NL) is a fully immersive virtual environment equipped with a six degree of freedom treadmill-motion platform. The CAREN provides an ideal setting to investigate gait biomechanics with the ability to simulate various stability-challenging tasks. Currently, there is no accepted standard for quantifying how stable or unstable an individual is during walking. The ability to quantify changes in dynamic stability during ambulation would provide a quantifiable metric to identify individuals with increased fall risk and to assess improvements in gait throughout a rehabilitation program.

1.2 Rationale

Researchers and clinicians have used the CAREN system for the last decade [16-18]; however, there is little published literature on the technical capabilities and limitations of this virtual environment. Research that evaluates and accurately describes the CAREN system technical characteristics is needed to develop standardized approaches for this virtual reality tool and to inform clinicians and researchers regarding potential errors associated with this virtual environment.

One of the CAREN system features is a self-paced treadmill mode. Unlike conventional treadmills, a self-paced treadmill automatically adjusts speed in real-time to match the user's walking velocity. Self-paced mode could potentially enable virtual environments with multiple movement scenarios to elicit more natural gait responses that are not possible with conventional treadmills. However, uncertainty exists regarding self-paced mode application and whether people with physical disabilities would utilize self-paced mode within complex virtual environments. Research on gait biomechanics that compares fixed speed and self-paced speed treadmill modes will help resolve this uncertainty.

Walking presents additional challenges for individuals with a lower limb amputation due to the loss of lower limb musculature and proprioception. While a reasonable understanding exists about transtibial amputee gait on level ground, limited information is available on amputee walking stability when walking over more challenging terrain. Virtual environments, consisting of a variety of stability-challenging terrain, should be an effective approach for assessing walking stability of transtibial amputees. Research is required to examine stability outcome measures for a range of walking activities available in a state-of-the-art virtual environment, such as the CAREN system.

1.3 Objectives and hypotheses

This thesis examines the CAREN virtual environment's technical capabilities and application of these capabilities to biomechanical research questions. The three objectives of this thesis are:

1) Evaluate technical aspects of the virtual environment by examining the motion platform, treadmill, and force plate characteristics to quantify capabilities of the CAREN-Extended system at The Ottawa Hospital Rehabilitation Centre.

   Hypotheses:

   a. Platform and treadmill operation are consistent with manufacture specifications.
   b. Platform movement affects force plate signals.
   c. Treadmill operation affects force plate signals.

2) Compare gait biomechanics between fixed and self-paced treadmill speeds for able-bodied (AB) and unilateral transtibial amputee (TT) populations to understand how people vary their walking speed in a stability-challenging virtual environment.

Hypotheses:

a. Gait biomechanics between fixed and self-paced treadmill speeds are not significantly different, when speed is considered as a covariate.

b. Able-bodied and transtibial amputees walk slower for stability-challenging terrain compared to level treadmill walking.

c. Transtibial amputees walk slower for stability-challenging terrain compared to able-bodied.

3) Assess stability performance for able-bodied and transtibial amputees when walking in a stability-challenging virtual environment.

Hypotheses:

a. Margin-of-stability, step strategies, and gait variability for each non-level condition are significantly difference from level treadmill walking, for able-bodied and transtibial amputee participants.

b. Transtibial amputees are significantly different from able-bodied for margin-of-stability, step strategies, and gait variability.

1.4 Contributions

The virtual environment evaluation in this thesis provided researchers and clinicians with a knowledge base of CAREN technical capabilities, including factors to consider when analyzing and interpreting data collected using this virtual environment. This information can also help with developing CAREN-based research and rehabilitation scenarios.

The research on self-paced treadmill mode demonstrated the importance of allowing a person to change their walking speed when attempting to assess natural gait, especially for a variety of non-level walking conditions. This was observed for both able-bodied and transtibial amputee populations. The research also identified the importance of training and acclimation time to ensure that people use this mode appropriately.

This research will also benefit those who use a transtibial prosthesis by providing a better understanding of how and when they are unstable when performing movement activities. This thesis identified outcome measures that are appropriate for dynamic stability assessment within a CAREN virtual scenario. Quantitative biomechanical outcomes from a walking stability assessment could guide therapy, prosthesis configuration, and new prosthetic technologies. These developments can positively affect a person's mobility in the community and thereby enhance their quality of life.

1.5 Thesis outline

This thesis is presented in paper (manuscript) format and is divided into six chapters. Chapter 2 reviews able-bodied and transtibial amputee gait for level, slope, and uneven terrain. Chapter 2 also provides a brief overview of current virtual environments and self-paced treadmill technologies. Chapter 2 concludes with a review of common walking stability concepts and outcome measures currently used to quantify stability during walking. Chapters 3-5 of this thesis contain three journal manuscripts that have been accepted or submitted for publication.

Chapter 3 contains an accepted journal manuscript that addresses objective 1 and presents an evaluation of the CAREN-Extended system's motion platform, treadmill, and force plates to document the virtual environment's capabilities and to identify factors that may affect data collected with this virtual environment.

Chapter 4 contains a submitted journal manuscript that addresses objective 2 and compares gait biomechanics between fixed and self-paced treadmill speeds for able-bodied and unilateral transtibial amputee participants. This chapter also examines factors that may affect the self-paced treadmill mode.

Chapter 5 addresses objective 3 and examines AB and TT participant walking strategies in a stability-challenging virtual environment, consisting of sloped and uneven terrain. This journal manuscript was also submitted for publication.

Chapter 6 presents a thesis summary, overview of the thesis contributions, and suggestions for future work.

Chapter 2: Literature Review

The purpose of this chapter is to introduce walking biomechanics terminology and to review able-bodied and transtibial amputee gait for level, sloped, and uneven surfaces. Three walking stability concepts are presented and commonly used walking stability outcome measures are also reviewed. The chapter concludes with a description of various virtual environments, including systems that can provide variable terrain and provide self-paced treadmill operation.

2.1 Able-bodied and transtibial amputee gait

Biomechanical analysis can identify deviations from able-bodied walking and provide information for developing interventions that improve walking. Walking can be divided into repetitive gait cycles, typically based on strides. The two primary phases of a gait cycle are stance and swing, and these phases are further sub divided into initial contact, early stance, mid stance, late stance, initial swing, mid swing, and terminal swing [19]. These phases are illustrated in Figure 2.1, definitions for each phase are provided in Table 2.1, and definitions of joint motions used to describe the body position during gait are illustrated in Figure 2.2. The gait phases described for level walking can be easily translated to other walking surfaces.

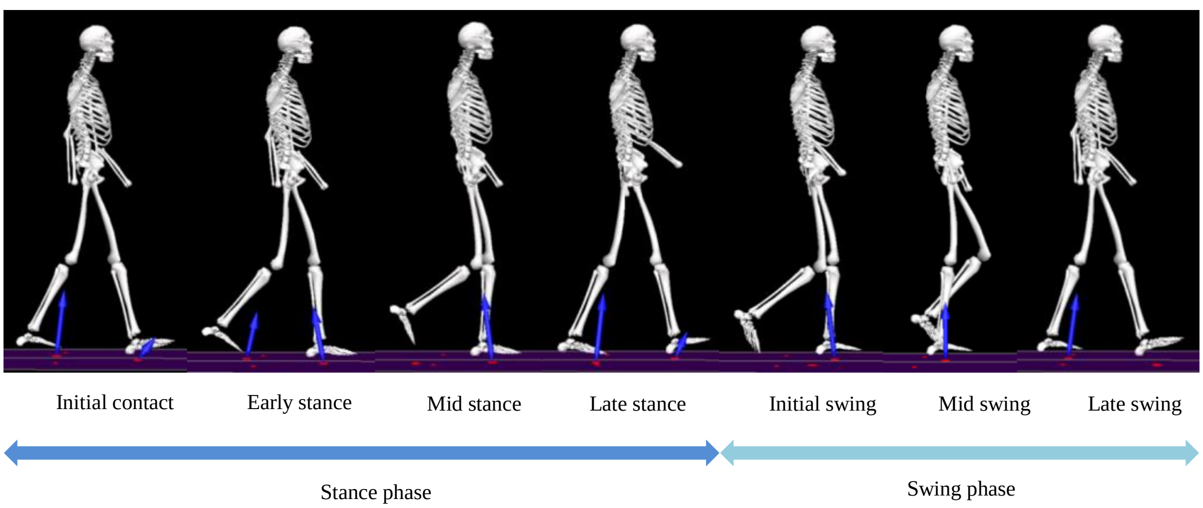

Figure 2.1. Phases of a single gait cycle during level ground walking from a sample walking trial. Description of each phase is provided in Table 2.1.

Table 2.1. Main phases of the gait cycle [19].

Gait phase	Definition
Stance phase	Phase with the limb in contact with the ground, beginning when the foot contacts the ground and ends when the foot leaves the ground.
Initial contact	First foot contact with the ground.
Early stance	Begins when the foot contacts the ground and the ends when the opposite limb leaves the ground (0-10% of the gait cycle). Also known as initial double support or loading response.
Mid stance	Begins when the opposite limb leaves the ground and ends when the opposite limb contacts the ground (10-50% of the gait cycle). Also known as single leg stance since the body is supported by one leg.
Late stance	Begins when the opposite limb contacts the ground and ends when the limb leaves the ground (50-60% of the gait cycle). Also known as terminal double support.
Swing	Phase with the limb in the air, beginning when the foot leaves the ground at toe-off and ends when the foot contacts the ground for the next gait cycle.
Initial swing	Begins at toe-off and ends when the limb is directly opposite the other limb (60-75% of the gait cycle).
Mid swing	When the lower leg is swung forward in front of the body (75-85% of the gait cycle).
Terminal swing	When the knee extends to prepare for the subsequent foot contact (85-100% of the gait cycle).

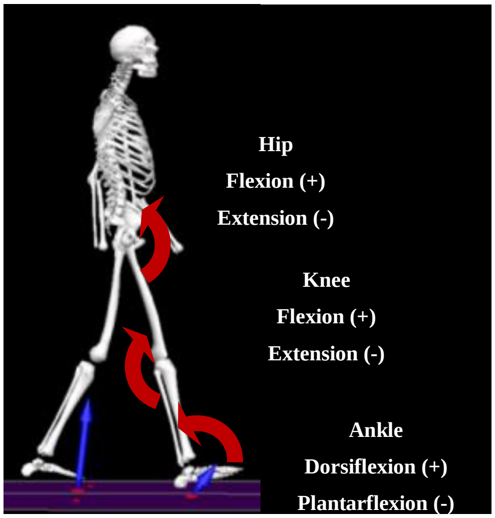

Figure 2.2. Ankle, knee, and hip joint motion in the sagittal plane.

2.1.1 Level walking

During level ground walking, able-bodied individuals use four main strategies to progress the body forward:

1) Large knee extensor moment during early stance to help prevent the knee from collapsing and keeping the body upright,
2) Controlled dorsiflexion motion and moment during stance to provide a smooth transition over the limb,
3) Plantarflexion moment and ankle power during late stance to push the body forward and away from the ground,
4) Ankle, knee, hip flexion during swing to ensure the limb clears the ground and to prevent tripping.

Additionally, the upper body remains close to neutral throughout the gait cycle, with some rotation toward the lead limb.

Current prosthetic devices, combined with rehabilitative training, allow transtibial amputees to walk similarly to able-bodied individuals over level ground. Passive prosthetic feet are able to mimic the motion and moment of able-bodied individuals, but are unable to produce the plantarflexion motion and produce ankle power during late stance [20]. Therefore, transtibial amputees typically display asymmetries between the prosthetic and intact limbs [21, 22]. The greatest deviations are observed at the ankle during late stance, demonstrated by the lack of prosthetic plantarflexion and reduced peak ankle power. Knee flexion during early stance is typically reduced, due to the inability to position the knee forward, over the foot, due to a lack of prosthetic dorsiflexion. Gait deviations such as hip hiking, limb vaulting, and limb circumduction can occur to provide limb clearance during swing phase.

2.1.2 Incline walking

As slope increases during uphill walking, able-bodied individuals contact the ground with more ankle dorsiflexion and more knee and hip flexion compared to level ground walking [23-25]. The trunk is generally positioned more anteriorly, to aid in forward propulsion [26], and range-of-motion at the ankle, knee, and hip is greater than level ground walking. Joint moment and power patterns are similar to level ground, but larger moments at the ankle, knee, and hip are required for larger inclines to raise the body uphill.

Although there is some literature investigating able-bodied gait strategies for incline walking, there is little literature for the transtibial amputee population. Transtibial amputees make similar postural adjustments during uphill walking compared to able-bodied individuals [27, 28]; however, transtibial amputees have a more extended knee during early stance. Reduced knee flexion during uphill walking may be due to limited ankle dorsiflexion of the prosthetic foot, which restricts the knee's ability to be positioned forward during early stance. Transtibial amputees also display similar deviations at the ankle, such as reduced prosthetic ankle plantarflexion and associated ankle moment and power during late stance.

2.1.3 Decline walking

Walking patterns during uphill walking are more similar to level ground than downhill walking. During level ground and uphill walking, the body progresses against gravity whereas gravity pulls the body down the slope during downhill walking. This results in a strategy change at the knee and hip to provide a controlled lowering of the body.

As the slope decreases, able-bodied individuals contact the ground with similar joint angles compared to level walking; however, lateral pelvis drop is greater, lowering the limb to contact the ground [23]. The trunk is positioned more posteriorly during decline relative to level walking, likely necessary to help decelerate the body [26]. Unilateral transtibial amputees typically contact the ground with a straighter leg on both prosthetic and intact limbs, demonstrated by more extended hip and knee angles [29]. At initial contact on the intact limb, the prosthetic limb is not able to reach the same amount of dorsiflexion as observed on an able-bodied ankle (i.e., late stance on prosthetic side), which restricts the body from lowering. Hence, the hip and knee on the intact side must extend to create a longer effective limb to reach the ground. At initial contact on the prosthetic side, the hip and knee are also more extended even though ankle dorsiflexion and knee flexion on the intact limb (i.e., late stance on intact side) are comparable to able-bodied. A more extended hip and knee suggests transtibial amputees took shorter steps on the prosthetic side. Shorter steps reduce the vertical distance the body is lowered for each step and, if the limb is positioned closer to the body, transfer weight onto this limb is easier [28]. However, similar to able-bodied, transtibial amputees may increase their pelvis drop during early stance, instead of taking shorter steps, as a strategy to increase the effective limb length and to help lower the body down the slope [28, 29].

Overall, the key differences between downhill walking and level walking are large knee power absorption as the limb is loaded and larger knee flexion maintained through stance to facilitate a controlled lowering of the body. Although the lack of ankle plantarflexion was the greatest limitation for transtibial amputees during level ground walking, the greatest limitation for uphill and downhill walking for transtibial amputees may be ankle dorsiflexion. Passive ankle-foot prostheses are not able to provide the range of dorsiflexion necessary throughout stance, thereby constraining range-of-motion at other joints.

2.1.4 Cross slope walking

A cross slope is the presence of a non-zero slope in the direction perpendicular to the direction of progression (Figure 2.3). Cross slope walking biomechanics has been investigated far less than level or incline/decline walking, even though it is a common real-world surface.

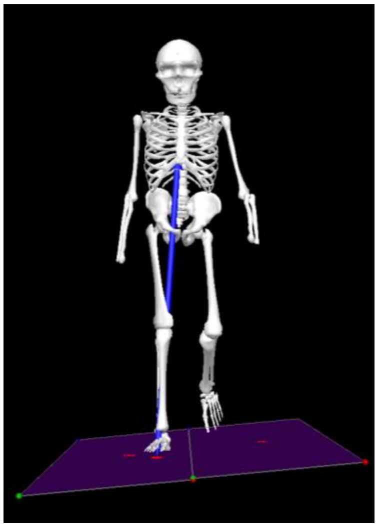

Figure 2.3. Cross slope walking surface.

Unlike level and incline/decline surfaces, walking over a cross slope requires an asymmetrical gait pattern to adapt to the height difference between the limbs while keeping the body vertical. Previous research reported significant asymmetries between limbs, even at a modest 6°cross slope [30]. The limb at the top of the cross slope must increase ankle, hip, and knee flexion to decrease the effective limb length, whereas the limb at the bottom of the cross slope must increase ankle, knee, and hip extension to increase the effective limb length [30]. Able-bodied subjects also increase frontal plane motion at the ankle [31] and pelvis [30] to adapt to these effective limb length discrepancies.

Currently, there is no biomechanical literature on cross-slope walking for lower extremity amputees. The asymmetries from walking over a cross slope are expected to pose a greater challenge to maintaining balance and forward progression for lower limb amputees.

2.1.5 Walking over uneven terrain

The uneven surfaces used to investigate gait biomechanics and stability are diverse, ranging from wooden blocks beneath a carpet, loose rocks, external perturbations via a motion platform or harness, and compliant surfaces [32-35]. Although the surfaces are different, the overall gait adaptations reported in these studies are similar.

When walking on uneven ground, able-bodied and transtibial amputee participants adopt a more cautious gait pattern characterized by shorter, wider, more variable steps and slower walking velocity, compared to walking over a level surface [33, 34, 36]. Other observations include increased ankle, knee, and hip flexion at foot contact, contributing to an overall lowering of the body to increase stability while walking over uneven surfaces [33, 34, 37, 38]. Studies also reported adaptive strategies to avoid the loss of balance, such as a decreased angle between the surface and the foot at foot contact [36, 39] and increased foot clearance during swing [36, 38, 40]. Transtibial amputee participants also exhibited larger outcome measure variability on their intact limb [41], which is likely a compensation for the loss of ankle control on the prosthetic limb.

2.2 Walking stability

In biomechanics, there has been considerable research on the mechanisms underlying stable human bipedal walking. However, there still is no universally accepted standard to quantify how stable or unstable an individual is during walking, and the mechanisms are only beginning to be understood. Stability can be defined as:

> "the property of a body that causes it, when disturbed from a condition of equilibrium or steady motion, to develop forces or moments that restore the original condition" [42].

Stable walking is achieved by incorporating feedback from the visual, vestibular, and somatosensory systems to generate appropriate responses to maintain forward progression of the body and upright posture without falling. The ability to quantify stability changes accurately when stability is challenged and to quantify gait strategies for maintaining stable gait would be valuable as an assessment tool and to develop interventions to improve walking stability. This section presents three approaches commonly used in walking stability research: 1) centre of mass versus base of support, 2) step strategies, and 3) gait variability.

2.2.1 Centre of mass versus base of support

Walking stability is often investigated by examining the relationship between the centre of mass (COM) and base of support (BOS) based on an inverted pendulum model [43]. A person is “stable” if the COM vertical projection remains within the BOS boundaries (defined by the feet on the support surface) and the person does not fall. A person is “unstable” if their COM travels beyond the BOS and the person falls. However, while this relationship between the COM and BOS works with static balance applications; COM and BOS are not sufficiently predictive for dynamic tasks, such as bipedal walking. Humans are able to produce active corrective responses by making postural adjustments, other gait strategies (e.g., foot placement), and corrective muscle control. Therefore, if the COM reaches or goes beyond the BOS, it is possible to regain stability dynamically by generating the appropriate corrective forces. However, if the necessary corrective action exceeds the individual’s ability to exert a sufficiently strong corrective response, a fall will still occur. A common method used to examine walking stability based on this approach is the margin-of-stability assessment. Margin-of-stability examines the minimum distance (margin) between the COM and BOS, accounting for the direction and speed of the body, to obtain a better estimate of dynamic stability.

2.2.2 Step strategies

Humans adjust gait both predicatively and responsively when walking stability is challenged. A number of compensatory gait strategies were identified in the literature as important for improving or enhancing stability. One strategy is a more cautious step pattern, characterized by shorter, wider, slower steps, and longer double support times [1, 35, 41, 44]. Step length and step width affect the size of the BOS, and double support time affects the duration of the most stable BOS of the gait cycle, when two feet are in contact with the support surface. This cautious step pattern is often observed for people with balance deficits when walking over a level surface, compared to a control group [1, 41, 44]. Other gait strategies are reduced angle between the foot and the surface at foot contact when anticipating a potential slip [33, 39, 45] and increased foot clearance during the swing phase when anticipating a potential trip [33, 40, 46, 47].

2.2.3 Gait variability

The smoothness or quality of gait patterns is a common stability indicator, and indirectly an indicator of fall risk. Gait variability assessment is often a statistical process where standard deviations or root-mean-squares of gait outcomes are the dependent measures, such as temporal-spatial parameters, joint kinematics, and kinetics. However, these assessments do not account for the inherent variability of the task or walking surface. When interpreting results, it is important to consider that variability outcome measure may reflect the various degrees of freedom available in order to adapt to a variety of terrain, which does not necessarily indicate instability [48, 49].

2.3 Walking stability measures

The following sections provide a detailed review of common stability measures for each walking stability approach:

2.3.1 Centre of mass versus base of support

2.3.1.1 Margin-of-stability

Margin-of-stability is based on the relationship between the COM and BOS during walking. This measure is an indicator of stability and examines the minimum difference between the COM and BOS in the medial-lateral axis or how close the COM travels to the medial-lateral boundary of the BOS (Figure 2.4). This occurs when the BOS transitions from double limb support to single limb support during early stance. However, simply examining the distance or margin between the COM and BOS overestimates the stability margin for dynamic tasks. Hof et al. proposed a method based on the COM and BOS relationship to obtain a better estimate of dynamic stability by incorporating a velocity component to account for the body's speed and direction [50, 51]. An offset based on the COM velocity is added to the COM (extrapolated COM), reducing the overall margin and resulting in a better margin of stability estimate. A known limitation of this approach is the assumption that walking is mathematically described using an inverted pendulum model, which doesn't account for active muscle control of the lower limb or movement of the arms and trunk [51].

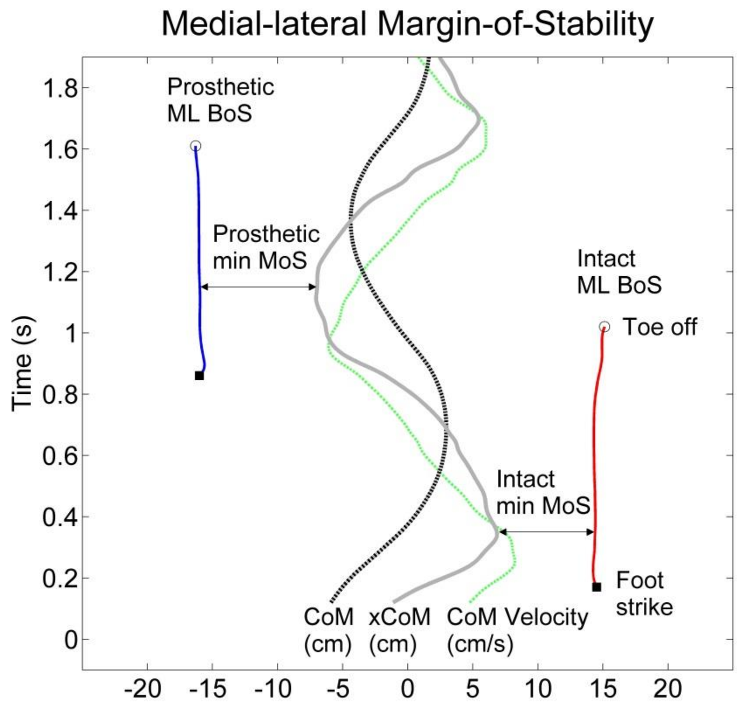

Figure 2.4. Medial-lateral (ML) margin-of-stability (MoS), where the minimum distance between the extrapolated center-of-mass (xCOM) and lateral base of support (BOS) is used as an indication of stability during walking.

In the literature, the boundary of the BOS is defined using one of two methods: centre of pressure obtained from force plate data or position of the foot markers(s). Hof et al., who proposed the margin-of-stability as an indicator of dynamic stability, reported normal margin-of-stability in the medial-lateral direction of 1-3 cm [52, 53] using centre of pressure to define the BOS boundary (Table 2.2). Day et al. also used centre of pressure to define the boundary of the BOS, but reported much larger margin-of-stability in the medial-lateral direction of approximately 8 cm for young able-bodied [54]. These values may be higher because the control group walked at a very slow speed (0.3m/s) to match the speed of the spinal cord injury patient group. Majority of research uses foot marker(s) to define the boundary of the BOS removing the need for force plate data. Reported margin-of-stability values for those studies range 5-10 cm (Table 2.2) for a young able-bodied population [12, 15, 55, 56], which is comparable to initial MOS values reported by Hof et al. when accounting for foot width. Table 2.2 shows the mean and mean of the standard deviations for medial-lateral margin-of-stability for level walking for various populations.

Table 2.2. Mean and mean of the standard deviations (mSD) for medial-lateral margin-of-stability for level walking. AB=able-bodied; TT=unilateral transtibial amputee; TF=unilateral transfemoral amputee; SCI=spinal cord injury; L=left limb; R=right limb; I=amputee intact limb; P=amputee prosthetic limb. Standard deviation is in brackets and NR refers to data not reported.

Author, Year	Pop.	Condition	Speed (m/s)	L,I Mean (cm)	R,P Mean (cm)	L,I mSD (cm)	R,P mSD (cm)
Hof, 2007	AB	Ground, slow	NR	1.40 (0.72)	1.35 (0.74)	0.325	0.319
		Ground, normal	NR	1.61 (0.71)	1.67 (0.70)	0.378	0.362
		Ground, fast	NR	1.81 (0.61)	1.86 (0.57)	0.403	0.363
Hof, 2010	AB	Treadmill	Froude3	2.1 (0.60)	2.2 (0.60)	0.6	0.5
Rosenblatt, 2010	AB	Treadmill	1.09 (0.17)	2.1 (0.60)	2.2 (0.60)	0.6	0.5
		Ground	1.41 (0.16)	6.66 (2.63)	6.88 (2.41)	NR	NR
Day, 2011	AB	Ground	0.30	8.30 (2.50)	NR	NR	NR
Curtze, 2011	AB	Ground	1.26 (0.13)	1.60 (5.17)	1.92 (3.10)	NR	NR
McAndrew, 2012	AB	Treadmill	NR	5.95 (1.41)	9.71 (1.66)	2.2 (0.72)	NR
Gates, 2013	AB	Ground	0.71 (0.03)	7.8 (0.74)	8.7 (0.76)	0.79 (0.21)	0.72 (0.13)
		Ground	0.95 (0.04)	8.2 (0.69)	8.6 (0.76)	0.66 (0.18)	0.80 (0.14)
		Ground	1.19 (0.05)	8.5 (0.65)	8.6 (0.65)	0.70 (0.13)	0.63 (0.15)
		Ground	1.42 (0.06)	8.7 (0.71)	8.9 (0.53)	0.63 (0.12)	0.63 (0.15)
Hak, 2013	AB	Treadmill	1.43 (0.17)	4.8 (0.95)	NR	NR	NR
Beltran, 2014	AB	Treadmill	Froude3	8.1 (0.53)	8.7 (0.51)	0.88 (0.05)	0.90 (0.04)
Curtze, 2011	TT	Ground	1.17 (0.13)	2.46 (1.47)	1.76 (2.37)	NR	NR
Gates, 2013	TT	Ground	0.73 (0.02)	8.8 (0.82)	7.7 (0.63)	0.71(0.18)	0.61 (0.15)
		Ground	0.97 (0.03)	8.8 (0.68)	7.9 (0.63)	0.57 (0.13)	0.54 (0.22)
		Ground	1.21 (0.03)	9.1 (0.66)	8.3 (0.58)	0.55 (0.18)	0.65 (0.18)
		Ground	1.45 (0.04)	9.1 (0.71)	8.5 (0.71)	0.47 (0.08)	0.52 (0.09)
Hak, 2013	TT	treadmill	1.27 (0.10)	6.3 (1.50)	NR	NR	NR
Beltran, 2014	TT	treadmill	Froude3	9.6 (0.31)	9.6 (0.36)	0.98 (0.09)	0.92 (0.08)
Hof, 2007	TF	ground, slow	NR	1.62 (0.42)	2.42 (0.36)	0.289	0.403
		ground, normal	NR	1.90 (0.62)	2.74 (0.54)	0.278	0.384
		ground, fast	NR	2.20 (0.92)	3.25 (0.88)	0.320	0.477
Day, 2011	SCI	ground	0.30	10.06 (3.2)	NR	NR	NR

Previous research demonstrated margin-of-stability increased with speed [15], when walking with surface perturbations [12, 55], and when walking with increased cognitive demand [52, 55]. Lower limb amputees also walked with wider margins than able-bodied [15, 32, 52]. Additionally, transtibial amputees walked with larger margin-of-stability on their intact limb compared to their prosthetic limb [15, 32], but Hof et al. reported transfemoral amputees walked with larger margin-of-stability on their prosthetic limb than their intact limb [52] and Beltran et al. reported no limb differences [57]. Studies that reported margin-of-stability values for right and left limbs showed able-bodied also display limb differences [32, 55], which may be related to limb dominance.

More recent studies reported margin-of-stability variability was greater when walking over loose rocks [15] and medial-lateral surface and visual perturbations [55], but did not change with increasing speed [15, 52] or with increased cognitive demand [52]. A summary of these values can be found in Table 2.3.

Table 2.3. Mean and mean of the standard deviations (mSD) for medial-lateral margin-of-stability for challenging walking tasks. AB=able-bodied; TT=unilateral transtibial amputee; TF=unilateral transfemoral amputee; SCI=spinal cord injury; L=left limb; R=right limb; I=amputee intact limb; P=amputee prosthetic limb. Standard deviation is in brackets and NR refers to data not reported.

Author, year	Pop.	Condition	Speed (m/s)	L,I Mean (cm)	R,P Mean (cm)	L,I mSD (cm)	R,P mSD (cm)
Hof, 2007	AB	Level, w/ stroop	NR	1.65 (0.60)	1.66 (0.57)	0.338	0.302
Hof, 2007	TF	Level, w/ stroop	NR	2.11 (0.69)	2.99 (0.46)	0.301	0.400
Curtze, 2011	AB	Custom irregular	1.21 (0.17)	1.36 (4.22)	2.32 (3.85)	NR	NR
McAndrew, 2012	AB	ML surface perturbations	NR	6.70 (3.58)	10.37 (3.73)	3.9 (0.63)	NR
McAndrew, 2012	AB	ML visual perturbations	NR	6.74 (4.07)	10.52 (3.98)	4.0 (1.47)	NR
Gates, 2012	AB	Rocks	0.71 (0.03)	8.0 (0.85)	8.4 (0.71)	0.97 (0.19)	1.11 (0.15)
	AB	Rocks	0.95 (0.04)	8.2 (0.88)	8.8 (0.69)	1.02 (0.22)	1.00 (0.16)
	AB	Rocks	1.19 (0.05)	8.3 (0.93)	8.8 (0.50)	1.03 (0.21)	1.04 (0.20)
	AB	Rocks	1.42 (0.06)	8.3 (0.71)	9.1 (0.49)	0.98 (0.20)	0.98 (0.17)
Hak, 2013	AB	ML surface perturbations	1.44 (0.17)	5.6 (0.67)	NR	NR	NR
Beltran, 2014	AB	ML surface perturbations	Froude3	8.5 (0.54)	9.2 (0.53)	1.84 (0.09)	1.86 (0.10)
	AB	ML visual perturbations	Froude3	8.4 (0.53)	9.0 (0.53)	1.74 (0.14)	1.74 (0.14)
Curtze, 2011	TT	Custom irregular	1.12 (0.21)	2.19 (1.29)	1.59 (4.1)	NR	NR
Gates, 2012	TT	Rocks	0.73 (0.02)	8.9 (0.83)	7.7 (0.82)	1.34 (0.30)	1.30 (0.60)
	TT	Rocks	0.97 (0.03)	9.0 (0.61)	8.3 (0.50)	0.95 (0.18)	1.01 (0.29)
	TT	Rocks	1.21 (0.03)	9.2 (0.61)	8.5 (0.44)	1.06 (0.25)	1.02 (0.32)
	TT	Rocks	1.45 (0.04)	9.1 (0.83)	8.9 (0.76)	0.99 (0.36)	1.04 (0.19)
Hak, 2013	TT	ML surface perturbations	1.25 (0.12)	7.0 (1.60)	NR	NR	NR
Beltran, 2014	TT	ML surface perturbations	Froude3	10.5 (0.56)	10.5 (0.65)	2.43 (0.26)	2.43 (0.31)
	TT	ML visual perturbations	Froude3	10.1 (0.37)	10.0 (0.48)	1.72 (0.21)	1.71 (0.22)

2.3.2 Step strategies

2.3.2.1 Temporal-spatial parameters

Temporal-spatial parameters describe the gait cycle in terms of time and distance. Commonly used parameters, their units of measurement and description are shown in Table 2.4. These parameters have been used to provide information on walking ability, which has been associated with walking stability [58]. Slower walking velocity is observed in many patient populations, which is thought to improve walking stability [1, 34, 49]. Slower walking velocities are accompanied by shorter step lengths, faster step times, and longer double support times [1, 35, 41, 44, 58]. This step strategy is referred to as a cautious or protective gait pattern. Longer double support times and larger step width increase the BOS size as well as the duration of the most stable phase of gait, enhancing stability. Furthermore, able-bodied participants adopt this cautious step pattern when walking over more challenging terrain [32, 34-36, 41, 45, 52, 59]. Table 2.5, Table 2.6, Table 2.7 present a temporal-spatial summary for level walking, slope walking, and for stability-challenging conditions, respectively.

Table 2.4. Temporal-spatial parameter description.

Parameter	**Description**
Walking velocity (m/s)	Average forward velocity over one gait cycle
Stride length (m)	Anterior-posterior distance over one gait cycle
Stride time (s)	Time to complete one gait cycle
Step length (m)	Anterior-posterior distance between toe-off and foot contact on the same limb
Step time (s)	Time between toe-off and foot contact on the same limb
Step width (m)	Medial-lateral distance between limbs at foot contact
Double support time (% of stride time)	Time between toe-off on one limb and foot contact on the opposite limb

Table 2.5. Temporal-spatial parameters for level walking. AB=able-bodied; TT=unilateral transtibial amputee; I=amputee intact limb; P=amputee prosthetic limb; DST=double support time. Standard deviation is in brackets and NR refers to data not reported.

Author, Year	Pop.	Condition	Walking speed (m/s)	Step Width (m)	Step Length (m)	Step Time (s)	Stride Length (m)	Stride Time (s)	DST (% stride)
Leroux, 2002	AB	Ground	1.03 (0.19)	NR	NR	NR	1.39 (0.19)	1.21 (0.15)	NR
Menz, 2003	AB	Ground	1.35 (0.19)	NR	0.73 (0.08)	0.58 (0.12)	NR	NR	NR
	Older	Ground	1.17 (0.16)	NR	0.65 (0.07)	0.56 (0.14)	NR	NR	NR
McIntosh, 2006	AB	Ground	1.57 (0.12)	NR	NR	NR	0.92 (0.05)	1.41 (0.08)	NR
Marigold, 2008	AB	Ground	1.33 (0.12)	0.11 (0.01)	0.72 (0.06)	NR	NR	NR	NR
	Older	Ground	1.20 (0.13)	0.10 (0.02)	0.63 (0.06)	NR	NR	NR	NR
Kendell, 2010	TT	Ground, I	NR	NR	NR	NR	NR	1.16 (0.07)	11.8 (9.0)
	TT	Ground, P	NR	NR	NR	NR	NR	1.17 (0.06)	13.3 (9.0)
Silverman, 2012	AB	Ground	NR	NR	0.67 (0.046)	NR	NR	NR	NR
Curtze, 2010	AB	Ground	1.26 (0.13)	0.07 (0.03)	0.72 (0.06)	0.57 (0.19)	NR	1.17 (0.08)	16.3 (2.4)
	TT	Ground	1.17 (0.13)	0.11 (0.03)	0.68 (0.07)	0.59 (0.18)	NR	1.20 (0.10)	17.4 (4.2)
Merryweather, 2011	AB	Ground	1.26 (0.12)	NR	NR	0.60 (0.07)	1.53 (0.12)	1.20 (0.05)	12.5 (1.7)
Gates, 2012	AB	Ground	0.71 (0.03)	0.10 (0.03)	0.53 (0.05)	0.73 (0.05)	NR	NR	NR
		Ground	0.95 (0.04)	0.10 (0.03)	0.61 (0.04)	0.64 (0.04)	NR	NR	NR
		Ground	1.19 (0.05)	0.10 (0.04)	0.68 (0.04)	0.58 (0.04)	NR	NR	NR
		Ground	1.42 (0.06)	0.11 (0.04)	0.75 (0.05)	0.54 (0.03)	NR	NR	NR
Gates, 2013	TT	Ground, I	0.73 (0.02)	0.12 (0.02)	0.58 (0.01)	0.77 (0.03)	NR	NR	NR
		Ground, I	0.97 (0.03)	0.11 (0.02)	0.64 (0.02)	0.67 (0.01)	NR	NR	NR
		Ground, I	1.21 (0.03)	0.10 (0.02)	0.71 (0.02)	0.60 (0.02)	NR	NR	NR
		Ground, I	1.45 (0.04)	0.10 (0.02)	0.78 (0.02)	0.55 (0.02)	NR	NR	NR
		Ground, P	0.73 (0.02)	0.12 (0.02)	0.55 (0.04)	0.75 (0.04)	NR	NR	NR
		Ground, P	0.97 (0.03)	0.10 (0.02)	0.64 (0.04)	0.65 (0.02)	NR	NR	NR
		Ground, P	1.21 (0.03)	0.11 (0.02)	0.71 (0.04)	0.59 (0.02)	NR	NR	NR
		Ground, P	1.45 (0.04)	0.11 (0.02)	0.80 (0.04)	0.55 (0.01)	NR	NR	NR
Hak, 2013	AB	Treadmill	1.43 (0.17)	0.20 (0.02)	0.67 (0.05)	0.50 (0.11)	NR	NR	NR
	TT	Treadmill	1.27 (0.10)	0.23 (0.02)	0.64 (0.05)	0.52 (0.08)	NR	NR	NR

Table 2.6. Temporal-spatial parameters for slope walking. AB=able-bodied; TT=unilateral transtibial amputee; I=amputee intact limb; P=amputee prosthetic limb; DST=double support time. Standard deviation is in brackets and NR refers to data not reported.

	Author, Year	Pop.	Slope	Walking speed (m/s)	Step Width (m)	Step Length (m)	Step Time (s)	Stride Length (m)	Stride Time (s)	DST (% stride)
INCLINE	Leroux, 2002	AB	+10°	1.03 (0.19)	NR	NR	NR	1.48 (0.19)	1.25 (0.18)	NR
	McIntosh, 2006	AB	+10°	1.68 (0.14)	NR	NR	NR	0.94 (0.06)	1.60 (0.14)	NR
			+8°	1.76 (0.19)	NR	NR	NR	0.88 (0.02)	1.62 (0.14)	NR
			+5°	1.73 (0.21)	NR	NR	NR	0.95 (0.03)	1.55 (0.08)	NR
	Vickers, 2008	AB	+5°	1.40 (0.13)	NR	NR	NR	1.45 (0.16)	1.03 (0.8)	NR
	Silverman, 2012	AB	+5°	NR	NR	0.71 (0.044)	NR	NR	NR	NR
			+10°	NR	NR	0.68 (0.063)	NR	NR	NR	NR
			+15°	NR	NR	0.64 (0.053)	NR	NR	NR	NR
	Vickers, 2008	TT	+5°	0.71 (0.15)	NR	NR	NR	0.98 (0.16)	1.39 (0.06)	NR
	Kendell, 2010	TT	+7°, I	NR	NR	NR	NR	NR	1.30 (0.14)	14.8 (1.6)
			+7°, P	NR	NR	NR	NR	NR	1.31 (0.14)	13.0 (2.1)
DECLINE	Leroux, 2002	AB	-10°	1.03 (0.19)	NR	NR	NR	1.25 (0.17)	1.13 (0.13)	NR
	McIntosh, 2006	AB	-10°	1.72 (0.19)	NR	NR	NR	0.88 (0.06)	1.50 (0.15)	NR
			-8°	1.72 (0.18)	NR	NR	NR	0.88 (0.05)	1.50 (0.14)	NR
			-5°	1.66 (0.14)	NR	NR	NR	0.89 (0.06)	1.47 (0.13)	NR
	Vickers, 2008	AB	-5°	1.42 (0.13)	NR	NR	NR	1.41 (0.18)	0.99 (0.08)	NR
	Silverman, 2012	AB	-5°	NR	NR	0.59 (0.053)	NR	NR	NR	NR
			-10°	NR	NR	0.62 (0.046)	NR	NR	NR	NR
			-15°	NR	NR	0.63 (0.031)	NR	NR	NR	NR
	Vickers, 2008	TT	-5°	0.66 (0.16)	NR	NR	NR	0.88 (0.21)	1.31 (0.08)	NR
	Kendell, 2010	TT	+7°, I	NR	NR	NR	NR	NR	1.21 (0.13)	13.8 (4.2)
			+7°, P	NR	NR	NR	NR	NR	1.22 (0.12)	10.3 (3.0)
CROSS SLOPE	Merryweather, 2011	AB	7°	1.25 (0.11)	NR	NR	0.60 (0.07)	1.52 (0.10)	1.20 (0.05)	12.1 (1.4)
			7° with large rocks	1.19 (0.12)	NR	NR	0.64 (0.10)	1.50 (0.11)	1.27 (0.08)	11.9 (2.1)

Table 2.7. Temporal-spatial parameters for uneven surfaces. AB=able-bodied; TT=unilateral transtibial amputee; I=amputee intact limb; P=amputee prosthetic limb; DST=double support time; ML=medial-lateral. Standard deviation is in brackets and NR refers to data not reported.

Author, Year	Pop.	Condition	Walking Speed (m/s)	Step Width (m)	Step Length (m)	Step Time (s)	Stride Length (m)	Stride Time (s)	DST (% stride)
Menz, 2003	AB	Custom irregular	1.34 (0.21)	NR	0.76 (0.08)	0.57 (0.14)	NR	NR	NR
	Older	Custom irregular	1.11 (0.18)	NR	0.66 (0.07)	0.59 (0.15)	NR	NR	NR
Marigold, 2008	AB	Custom irregular	1.37 (0.19)	0.12 (0.02)	0.76 (0.11)	NR	NR	NR	NR
	Older	Custom irregular	1.07 (0.19)	0.13 (0.03)	0.58 (0.11)	NR	NR	NR	NR
Kendell, 2010	TT	Foam mats, I	NR	NR	NR	NR	NR	1.34 (0.12)	14.8 (2.1)
	TT	Foam mats, P	NR	NR	NR	NR	NR	1.35 (0.10)	13.3 (2.2)
Curtze, 2010	AB	Custom irregular	1.21 (0.17)	0.08 (0.03)	0.71 (0.09)	0.58 (0.20)	1.18 (0.09)	NR	16.7 (2.4)
	TT	Custom irregular	1.12 (0.21)	0.11 (0.03)	0.66 (0.10)	0.60 (0.23)	1.21 (0.13)	NR	15.6 (2.2)
Merryweather, 2011	AB	Large rocks	1.18 (0.08)	NR	NR	0.62 (0.08)	1.47 (0.10)	1.24 (0.06)	12.5 (1.6)
Gates, 2012	AB	Rocks	0.71 (0.03)	0.11 (0.04)	0.53 (0.05)	0.72 (0.06)	NR	NR	NR
		Rocks	0.95 (0.04)	0.11 (0.03)	0.60 (0.06)	0.64 (0.05)	NR	NR	NR
		Rocks	1.19 (0.05)	0.11 (0.04)	0.68 (0.06)	0.58 (0.03)	NR	NR	NR
		Rocks	1.42 (0.06)	0.11 (0.03)	0.75 (0.07)	0.53 (0.04)	NR	NR	NR
Gates, 2012	TT	Rocks, I	0.73 (0.02)	0.13 (0.03)	0.56 (0.02)	0.74 (0.03)	NR	NR	NR
		Rocks, I	0.97 (0.03)	0.13 (0.03)	0.63 (0.02)	0.66 (0.02)	NR	NR	NR
		Rocks, I	1.21 (0.03)	0.12 (0.02)	0.68 (0.02)	0.59 (0.02)	NR	NR	NR
		Rocks, I	1.45 (0.04)	0.12 (0.03)	0.77 (0.02)	0.55 (0.02)	NR	NR	NR
		Rocks, P	0.73 (0.02)	0.13 (0.03)	0.54 (0.03)	0.75 (0.03)	NR	NR	NR
		Rocks, P	0.97 (0.03)	0.12 (0.02)	0.63 (0.04)	0.65 (0.03)	NR	NR	NR
		Rocks, P	1.21 (0.03)	0.12 (0.02)	0.72 (0.04)	0.58 (0.03)	NR	NR	NR
		Rocks, P	1.45 (0.04)	0.12 (0.02)	0.80 (0.03)	0.55 (0.02)	NR	NR	NR
Hak, 2013	AB	ML surface perturbations	1.44 (0.17)	0.23 (0.03)	0.70 (0.05)	0.48 (0.10)	NR	NR	NR
	TT	ML surface perturbations	1.25 (0.12)	0.26 (0.03)	0.61 (0.06)	0.51 (0.09)	NR	NR	NR

2.3.2.2 Foot angle

Foot kinematics at initial surface contact is an important predictor of falling risk due to a slip [60]. A slip occurs when the forces generated by body deceleration and initial loading are greater than the friction provided at the shoe-surface interface [60]. The shoe-surface interface can be affected by environmental factors (e.g., dry sidewalk or icy sidewalk), footwear (e.g., material properties), and human factors (e.g., gait deviations). Slip events can be characterized by a reduction in shear and vertical ground reaction forces and smaller forward progression of the centre of pressure under the foot (centre of pressure remains near the ankle joint).

Slip events can also be characterized by foot kinematics; including, higher impact heel velocity, slower foot angular velocity at foot contact, and faster forward velocity after foot contact [60]. Furthermore, Cham et al. [39] found that able-bodied participants adopted a different gait pattern when anticipating challenging surface conditions, such as a reduced angle between the foot and the surface at foot contact. Smaller foot-surface angles reduce the shear ground reaction forces, therefore reducing the required friction during walking and the potential risk of falling from a slip [33, 36, 39, 45].

Subsequent research has demonstrated foot-surface angles from 19-28 degrees for young able-bodied participants when walking over a level, dry surface (Table 2.8). Older able-bodied and unilateral transtibial amputees contacted the surface with a smaller foot-surface angle (i.e., flatter foot) compared to their young able-bodied control group [33, 45]. Young able-bodied [36, 39], older able-bodied [45], and unilateral transtibial amputee [33] also contacted the surface with a smaller foot-surface angle when walking over more challenging terrain or slippery surfaces [33, 39, 45] [33, 39, 45]. A summary of foot-surface angle measurements for various populations as found in the literature are shown in Table 2.8.

Table 2.8. Foot-surface angle (angle between the foot segment and the surface at foot contact) in the literature. AB=able-bodied; TT=unilateral transtibial amputee; I=amputee intact limb; P=amputee prosthetic limb. Standard deviation is in brackets and NR refers to data not reported.

Author, Year	Pop.	Condition	Speed (m/s)	Step length (m)	Foot-surface angle (°)
Cham, 2002	AB	Level ground (dry surface)	NR	NR	19.31 (0.06)
		+5° (dry surface)	NR	NR	21.99 (0.71)
		+10° (dry surface)	NR	NR	21.05 (0.7)
Cham, 2002	AB	Level ground (vinyl surface)	NR	NR	23.5 (3.7)
		+5° (vinyl surface)	NR	NR	26.4 (3.5)
		+10° (vinyl surface)	NR	NR	26.9 (4.9)
Menant, 2009	AB	Level, irregular, wet	1.30 (0.13)	0.703 (0.05)	23.5 (2.7)
	Older	Level, irregular, wet	0.93 (0.22)	0.546 (0.08)	17.5 (5)
	AB and older combined	Level	1.17 (0.18)	0.638 (0.06)	24.1 (3.3)
		Irregular	1.06 (0.18)	0.621 (0.07)	NR
		Wet	1.12 (0.15)	0.615 (0.06)	19.7 (4.2)
Gates, 2012	AB	Level ground	0.71 (0.03)	0.53 (0.05)	20.3 (3.8)
		Level ground	0.95 (0.04)	0.61 (0.04)	23.2 (3.3)
		Level ground	1.19 (0.05)	0.68 (0.05)	25.6 (2.5)
		Level ground	1.42 (0.06)	0.75 (0.05)	28.2 (2.5)
		Rocks	0.71 (0.03)	0.53 (0.05)	10.7(4.7)
		Rocks	0.95 (0.04)	0.60 (0.06)	12.8 (5.7)
		Rocks	1.19 (0.05)	0.68 (0.06)	15.3 (4.7)
		Rocks	1.42 (0.06)	0.75 (0.07)	16.5 (4.7)
Gates, 2012	TT	Level ground, I	0.73 (0.02)	0.58 (0.01)	21.8 (4.1)
		Level ground, I	0.97 (0.03)	0.64 (0.02)	23.6 (3.1)
		Level ground, I	1.21 (0.03)	0.71 (0.02)	25.6 (2.6)
		Level ground, I	1.45 (0.04)	0.78 (0.02)	28.5 (3.3)
		Level ground, P	0.73 (0.02)	0.55 (0.04)	23.2 (2.5)
		Level ground, P	0.97 (0.03)	0.64 (0.04)	25.8 (2.2)
		Level ground, P	1.21 (0.03)	0.71 (0.04)	28.0 (2.3)
		Level ground, P	1.45 (0.04)	0.80 (0.04)	29.9 (2.4)
		Rocks, I	0.73 (0.02)	0.56 (0.02)	12.0 (2.9)
		Rocks, I	0.97 (0.03)	0.63 (0.02)	12.6 (2.8)
		Rocks, I	1.21 (0.03)	0.68 (0.02)	15.0 (2.8)
		Rocks, I	1.45 (0.04)	0.77 (0.02)	18.1 (3.3)
		Rocks, P	0.73 (0.02)	0.54 (0.03)	18.7 (2.3)
		Rocks, P	0.97 (0.03)	0.63 (0.04)	21.5 (3.0)
		Rocks, P	1.21 (0.03)	0.72 (0.04)	21.9 (2.9)
		Rocks, P	1.45 (0.04)	0.80 (0.03)	24.7 (3.1)

2.3.2.3 Foot clearance

Foot kinematics during swing is an important factor for fall risk, with the distance between the sole and the walking surface related to tripping risk. During swing phase, the distance from the hip joint centre to the foot (effective limb length) must decrease to clear potential obstacles. Minimum foot clearance (MinFC) is typically 1 - 2 cm and is observed during mid-swing, when the foot is closest to the ground (Table 2.9).

Foot clearance is achieved by reducing effective limb length during swing via coordinated hip, knee, and ankle flexion. Moosabhoy et al. [61] demonstrated that foot clearance was most sensitive to ankle dorsiflexion, compared to hip and knee flexion. Physical issues that affect lower limb kinematics, such as lack of normal ankle dorsiflexion in lower limb prosthesis users, often results in compensatory kinematic responses to successfully provide foot clearance, such as limb circumduction, limb vaulting, lateral trunk lean, and/or hip hiking. These compensations attempt to decrease effective limb length or increase the distance between the hip joint centre and the ground by raising the upper body.

Foot clearance has been researched to determine relationships between people at risk of falling and minimum foot clearance. Older people walked with a smaller foot clearance than a young able-bodied population, suggesting that a lower foot clearance increases tripping risk [58, 62-64]. However, other studies reported that older [64, 65] and lower limb amputees [64] walked with a larger foot clearance than young able-bodied, suggesting that older adults actively increase foot clearance as an anticipatory strategy. Furthermore, Khandoker et al. [63] also demonstrated that older fallers (participants who fell at least once in the previous year) walked with larger foot clearance than older non-fallers (participants who had not fallen in the previous year). Conversely, young K-level 4 transtibial amputees only increased foot clearance on their intact limb, compared to able-bodied [33]. Other studies also demonstrated a foot clearance increase strategy as an adaptation during walking over loose rocks [33, 36, 59], uneven ground [40], slopes [40, 59, 63], and when vision was partially occluded [47, 66]. Table 2.9 summarizes minimum foot clearance values reported for level walking and Table 2.10 summarizes minimum foot clearance values reported for uneven surfaces.

Table 2.9. Minimum foot clearance for level walking. AB=able-bodied; TT=unilateral transtibial amputee; L=left limb; R=right limb; I=amputee intact limb; P=amputee prosthetic limb; mSD=mean of the standard deviations; IRQ=interquartile range. Standard deviation is in brackets and NR refers to data not reported.

Author, Year	Pop.	Condition	Speed (m/s)	Mean (cm)	mSD (cm)	Median (cm)	IRQ (cm)
Winter, 1990	AB	Ground	1.4	1.27 (0.59)	0.45 (0.31)	NR	NR
Moosabhoy, 2006	AB	Ground	1.4 (0.15)	1.90 (0.50)	NR	NR	NR
MacLellan, 2006	AB	Ground (N+4)	NR	2.27 (0.31)	NR	NR	NR
Begg, 2007	AB	Treadmill	1.15 (0.21)	1.55 (0.63)	0.25 (0.05)	1.50 (0.62)	0.31 (0.06)
Khandoker, 2008	AB	Treadmill	1.30 (0.16)	1.46 (0.52)	0.25 (0.05)	1.44 (0.52)	0.32 (0.08)
Graci, 2009	AB	Ground	1.24 (0.19)	1.96	0.86	1.85	0.88
Khandoker, 2010	AB	Treadmill	0.98	1.06 (0.56)	0.31 (0.11)	1.03 (0.55)	0.37 (0.15)
Mills, 2008	AB	Treadmill	1.12 (0.08)	NR	NR	1.49 (0.16)	0.43 (0.09)
Sparrow, 2008	AB	Treadmill, L	1.29	1.05	0.50	1.01	0.69
		Treadmill, R	1.29	1.44	0.34	1.41	0.43
Thies, 2011	AB	Ground	1.42 (0.19)	1.40 (0.70)	NR	NR	0.8 (0.60)
Schulz, 2011	AB	Ground	1.02 (0.23)*	0.85 (0.50)	NR	NR	NR
		Ground	1.5 (0.26)*	1.02 (0.45)	NR	NR	NR
		Ground	2.56 (0.49)*	1.46 (0.60)	NR	NR	NR
Merryweather, 2011	AB	Ground					
Gates, 2012	AB	Ground	0.71 (0.03)	1.76 (0.48)	NR	NR	NR
		Ground	0.95 (0.04)	1.82 (0.52)	NR	NR	NR
		Ground	1.19 (0.05)	2.05 (0.62)	NR	NR	NR
		Ground	1.42 (0.06)	2.12 (0.50)	NR	NR	NR
Winter, 1990	Older	Ground	NR	1.11 (0.53)	0.35 (0.20)	NR	NR
Begg, 2007	Older	Treadmill	0.88 (0.19)	1.43 (0.76)	0.32 (0.10)	1.41 (0.78)	0.40 (0.12)
Khandoker, 2008	Older	Treadmill	1.29 (0.16)	1.25 (0.47)	0.32 (0.09)	1.21 (0.47)	0.39 (0.11)
Mills, 2008	Older	Treadmill	1.12 (0.09)	NR	NR	1.38 (0.21)	0.53 (0.12)
Sparrow, 2008	Older	Treadmill, L	0.93	1.22	0.51	1.16	0.69
		Treadmill, R	0.93	1.97	0.46	1.96	0.60
Khandoker, 2008	Older fallers	Treadmill	0.91 (0.19)	2.02 (0.51)	0.47 (0.16)	2.00 (0.51)	0.62 (0.22)
Gates, 2012	TT	Ground, I	0.73 (0.02)	2.01 (0.51)	NR	NR	NR
		Ground, I	0.97 (0.03)	2.18 (0.58)	NR	NR	NR
		Ground, I	1.21 (0.03)	2.22 (0.54)	NR	NR	NR
		Ground, I	1.45 (0.04)	2.26 (0.54)	NR	NR	NR
		Ground, P	0.73 (0.02)	1.23 (0.42)	NR	NR	NR
		Ground, P	0.97 (0.03)	1.35 (0.65)	NR	NR	NR
		Ground, P	1.21 (0.03)	1.50 (0.45)	NR	NR	NR
		Ground, P	1.45 (0.04)	1.68 (0.58)	NR	NR	NR

* % of leg length/s

Table 2.10. Minimum foot clearance for slope and uneven ground walking, and walking with visual challenges. AB=able-bodied; TT=unilateral transtibial amputee; I=amputee intact limb; P=amputee prosthetic limb; mSD=mean of the standard deviations; IRQ=interquartile range. Standard deviation is in brackets and NR refers to data not reported.

	Author, Year	**Pop.**	**Condition**	**Speed (m/s)**	**Mean (cm)**	**mSD (cm)**	**Median (cm)**	**IRQ (cm)**
SLOPE	Khandoker, 2010	AB	Treadmill,+3°	0.98	0.79 (0.36)	0.24 (0.06)	0.77 (0.37)	0.30 (0.05)
	Thies, 2011	AB	Ground,+4.5°	1.41 (0.18)	2.8 (1.0)	NR	NR	0.5 (0.2)
	Khandoker, 2010	Older	Treadmill,+3°	0.87	0.91 (0.38)	0.11 (0.03)	0.88 (0.40)	0.37 (0.12)
	Khandoker, 2010	AB	Treadmill,-3°	0.98	1.37 (0.71)	0.38 (0.16)	1.36 (0.72)	0.48 (0.25)
	Thies, 2011	AB	Ground,-4.5°	1.42 (0.19)	3.2 (1.0)	NR	NR	0.5 (0.2)
	Khandoker, 2010	Older	Treadmill,-3°	0.87	1.13 (0.39)	0.12 (0.03)	1.09 (0.42)	0.42 (0.11)
	MacLellan, 2006	AB	Ground (N+4)	NR	8.78 (1.83)	NR	NR	NR
	Thies, 2011	AB	Ground, tactile	1.41 (0.20)	1.6 (0.7)	NR	NR	0.8 (0.4)
	Merryweather, 2011	AB	Rocks					
UNEVEN	Gates, 2012	AB	Rocks	0.71 (0.03)	6.33 (0.83)	NR	NR	NR
			Rocks	0.95 (0.04)	7.02 (0.98)	NR	NR	NR
			Rocks	1.19 (0.05)	7.71 (0.94)	NR	NR	NR
			Rocks	1.42 (0.06)	8.51 (0.87)	NR	NR	NR
	Gates, 2012	TT	Rocks, I	0.73 (0.02)	7.22 (0.98)	NR	NR	NR
			Rocks, I	0.97 (0.03)	7.88 (0.85)	NR	NR	NR
			Rocks, I	1.21 (0.03)	9.08 (0.94)	NR	NR	NR
			Rocks, I	1.45 (0.04)	8.90 (0.89)	NR	NR	NR
			Rocks, P	0.73 (0.02)	5.88 (0.72)	NR	NR	NR
			Rocks, P	0.97 (0.03)	6.30 (0.42)	NR	NR	NR
			Rocks, P	1.21 (0.03)	7.07 (0.49)	NR	NR	NR
			Rocks, P	1.45 (0.04)	7.40 (0.45)	NR	NR	NR
	Schulz, 2011	AB	Visible obstacles	1.02 (0.28)*	1.85 (0.51)	NR	NR	NR
			Visible obstacles	1.44 (0.23)*	2.22 (0.55)	NR	NR	NR
			Visible obstacles	2.41 (0.42)*	2.59 (0.92)	NR	NR	NR
			Hidden obstacles	1.06 (0.24)*	2.35 (0.51)	NR	NR	NR
			Hidden obstacles	1.47 (0.23)*	2.64 (0.58)	NR	NR	NR
			Hidden obstacles	2.31 (0.39)*	3.04 (0.85)	NR	NR	NR
VISUAL CUES	Graci, 2009	AB	Level w/ lower occlusion	NR	2.01	1.01	1.86	0.86
			Level w/ upper occlusion	NR	2.05	1.28	1.85	0.91
			Level w/ circumferential peripheral occlusion	1.20 (0.21)	2.18	1.29	1.92	1.06

* % of leg length/s

2.3.3 Gait variability

2.3.3.1 Step variability

Although the magnitude of temporal spatial parameters, foot angle, foot clearance, and margin-of-stability provides a step pattern assessment, variability of these parameters are also greater for populations with balance deficits, such as older and lower limb amputees, or when walking over more challenging surfaces [15, 33, 34, 67-70]. Gait parameter variability may be a stronger indicator of step-to-step changes when stability is challenged and the likelihood of falling is greater [55, 67-69, 71]. Table 2.11 and Table 2.12 summarize temporal-spatial variability values reported in the literature for various terrain and populations.

Table 2.11. Mean of standard deviation (mSD) for step width, step length, step time, and stride time for level walking. AB=able-bodied; TT=unilateral transtibial amputee; I=amputee intact limb; P=amputee prosthetic limb; ML=medial-lateral. Standard deviation is in brackets and NR refers to data not reported.

	Author	**Pop.**	**Condition**	**Step width (cm)**	**Step length (cm)**	**Step time (ms)**	**Stride time (s)**
LEVEL	Menz, 2003	AB	Ground	NR	NR	NR	0.03 (0.02)
	Marigold, 2008	AB	Ground	3.0 (1.0)	6.0 (0.7)	NR	NR
	Callisaya, 2010	Older male	Ground	2.21 (0.75)	2.81 (0.95)	2 (1)	NR
		Older female	Ground	1.99 (0.59)	2.61 (0.86)	2 (1)	NR
	Terry, 2012	AB	Treadmill	3.03 (0.57)	NR	NR	0.016 (0.003)
	Gates, 2012	AB	Ground, speed 3	2.0 (0.8)	2.0 (0.9)	1.2 (0.4)	NR
	Gates, 2013	TT	Ground, speed 1, I	3.4 (0.8)	2.0 (0.6)	2.4 (0.6)	NR
			Ground, speed 2, I	2.7 (0.7)	1.8 (0.6)	1.5 (0.3)	NR
			Ground, speed 3, I	2.9 (0.6)	1.6 (0.4)	1.0 (0.3)	NR
			Ground, speed 4, I	2.8 (0.7)	1.3 (0.3)	1.0 (0.3)	NR
			Ground, speed 1, P	2.2 (0.5)	2.1 (0.9)	2.5 (0.7)	NR
			Ground, speed 2, P	1.9 (0.4)	1.7 (0.4)	1.5 (0.3)	NR
			Ground, speed 3, P	2.4 (0.5)	1.5 (0.5)	1.2 (0.3)	NR
			Ground, speed 4, P	2.6 (0.6)	1.7 (0.4)	1.2 (0.2)	NR
	Menz, 2003	Older	Ground	NR	NR	NR	0.05 (0.02)
	Marigold, 2008	Older	Ground	3.0 (1.0)	9.0 (6.0)	NR	NR

Table 2.12. Mean of standard deviation (mSD) for step width, step length, step time, and stride time for stability-challenging terrain. AB=able-bodied; TT=unilateral transtibial amputee; I=amputee intact limb; P=amputee prosthetic limb; ML=medial-lateral. Standard deviation is in brackets and NR refers to data not reported.

	Author	Pop.	Condition	Step width (cm)	Step length (cm)	Step time (ms)	Stride time (s)
UNEVEN	Menz, 2003	AB	Custom irregular	NR	NR	NR	0.04 (0.02)
		Older	Custom irregular	NR	NR	NR	0.05 (0.01)
	Marigold, 2008	AB	Custom irregular (Natural)	6.0 (3.0)	12.0 (6.0)	NR	NR
		Older	Custom irregular (Natural)	7.0 (2.0)	12.0 (4.0)	NR	NR
	Terry, 2012	AB	Treadmill, ML perturbations (P1)	7.09 (1.2)	NR	NR	0.02 (0.003)
	Gates, 2012	AB	Rocks, speed 3	3.4 (1.3)	3.5 (0.1)	2.4 (0.8)	NR
	Gates, 2013	TT	Rocks, speed 1, I	6.5 (2.1)	3.9 (1.0)	4.0 (1.0)	NR
			Rocks, speed 2, I	5.9 (1.7)	4.2 (1.2)	2.9 (0.6)	NR
			Rocks, speed 3, I	5.8 (1.2)	4.9 (1.5)	2.4 (0.6)	NR
			Rocks, speed 4, I	4.4 (1.2)	3.6 (0.7)	2.0 (0.6)	NR
			Rocks, speed 1, P	4.3 (1.3)	4.4 (0.8)	2.9 (1.0)	NR
			Rocks, speed 2, P	3.7 (1.1)	3.3 (0.7)	3.5 (0.7)	NR
			Rocks, speed 3, P	4.0 (0.7)	4.5 (1.6)	2.1 (0.7)	NR
			Rocks, speed 4, P	5.1 (1.4)	4.0 (1.4)	2.6 (1.0)	NR

2.3.3.2 Trunk acceleration variability

During walking, the body's COM progresses through a cyclical and regular pattern of acceleration and deceleration [72]. Deviations in the timing and magnitude of the muscle activations are observed as a deviation from this pattern. Previous work suggests that acceleration-based analysis is capable of predicting minute changes in stability that in turn can lead to identifying individuals with an increased fall risk [72, 73]. Many acceleration-based stability analyses directly measure acceleration using an accelerometer attached to the trunk. The root-mean-square (RMS) is calculated to quantify trunk acceleration magnitude, where larger RMS values indicate less smoothness or more variability in the walking pattern [73]. Using this method, values reported over a level surface for able-bodied walking at a self-selected pace range from 0.11-0.14g in the medial-lateral direction, 0.15-0.18g in the anterior-posterior direction, and 0.18-0.23g in the vertical direction, in units of gravity. Details of measurements reported in the literature are shown in Table 2.13. Alternatively, acceleration can be derived from motion capture data. Marigold et al. [34] examined RMS from filtered segment data and reported slightly smaller RMS values (Table 2.13).

RMS analysis has identified differences between groups; such as, young and older [1, 34], older non-fallers and older fallers [74, 75], able-bodied and transfemoral amputees [13], or diabetic peripheral neuropathy [76]. Previous research found similar RMS values for healthy older and young able-bodied participants; hence, healthy older individuals may retain uncompromised ability to regulate body accelerations during gait [77] and, according to this type of analysis, are equally stable compared to young able-bodied population.

RMS also increased when walking over uneven surfaces, especially in the medial-lateral direction [13, 34]. However, Chang et al. [78] found that able-bodied individuals attenuated trunk accelerations when walking over foam mats. Marigold et al. [34] found an increase in trunk RMS when walking over multi-terrain irregular surface, which may be considered more challenging than foam mats.

Trunk acceleration RMS is also related to walking speed, with faster speeds resulting in larger RMS [73, 79, 80]. Although the majority of affected groups walked slower, studies that did not normalize RMS by speed found no significance between participant groups (even though they

Table 2.13. Trunk acceleration root-mean-square, in units of gravity (g). AB=able-bodied; TT=unilateral transtibial amputee; TF=unilateral transfemoral amputee; ML=medial-lateral; AP=anterior-posterior; VT=vertical. Standard deviation is in brackets and NR refers to data not reported.

	Author	**Pop.**	**Condition**	**Speed (m/s)**	**ML Mean (g)**	**AP Mean (g)**	**VT Mean (g)**
LEVEL	Kavanagh, 2004	AB	Level surface*	1.28 (0.15)	0.11 (0.08)	0.15 (0.10)	0.18 (0.12)
	Marigold, 2008	AB	Level surface*	1.33 (0.12)	0.06 (0.01)	0.10 (0.01)	0.17 (0.02)
	Kavanagh, 2009	AB	Level surface, slow	0.93 (0.11)	0.089 (0.005)	0.11 (0.005)	0.11 (0.005)
	Kavanagh, 2009	AB	Level surface, normal	1.32 (0.18)	0.13 (0.006)	0.17 (0.005)	0.20 (0.005)
	Kavanagh, 2009	AB	Level surface, fast	1.78 (0.29)	0.22 (0.01)	0.26 (0.01)	0.35 (0.01)
	Lamoth, 2010	AB	Level surface	1.41 (0.15)	0.14 (0.01)	0.14 (0.02)	NR
	Chang, 2010	AB	Level surface	1.27 (0.21)	0.131 (0.044)	0.176 (0.027)	0.232 (0.065)
	Kavanagh, 2004	Older	Level surface*	1.23 (0.15)	0.11 (0.07)	0.14 (0.10)	0.18 (0.11)
	Marigold, 2008	Older	Level surface*	1.20 (0.13)	0.08 (0.01)	0.11 (0.02)	0.16 (0.02)
	Senden, 2012	Older	Level surface	1.23 (0.22)	NR	NR	0.25 (0.07)
	Senden, 2012	Older fallers	Level surface	0.86 (0.26)	NR	NR	0.16 (0.07)
	Lamoth, 2010	TF	Level surface	1.27 (0.22)	0.21 (0.02)	0.17 (0.02)	NR
UNEVEN	Marigold, 2008	AB	Custom irregular*	1.37 (0.19)	0.07 (0.01)	0.13 (0.02)	0.19 (0.01)
	Lamoth, 2010	AB	Rough paved terrain	1.49 (0.15)	0.17 (0.01)	0.16 (0.02)	NR
	Chang, 2010	AB	Compliant surface	1.24 (0.22)	0.119 (0.032)	0.191 (0.026)	0.219 (0.056)
	Marigold, 2008	Older	Custom irregular*	1.07 (0.19)	0.10 (0.02)	0.13 (0.02)	0.17 (0.01)
	Lamoth, 2010	TF	Rough paved terrain	1.27 (0.22)	0.22 (0.02)	0.18 (0.01)	NR
COGNI-TIVE	Lamoth, 2010	AB	Level surface, dual task	1.33 (0.19)	0.13 (0.01)	0.13 (0.02)	NR
	Lamoth, 2010	TF	Level surface, dual task	1.18 (0.18)	0.20 (0.02)	0.16 (0.02)	NR

* normalized by speed

walked slower). These studies theorize that participants reduced walking speed and altered stepping strategy to maintain accelerations to a tolerable level [1]. Additionally, larger trunk accelerations may increase the likelihood of reaching the BOS boundary, leading to a potential fall.

2.4 Virtual environments

Virtual environments are emerging as a versatile tool for researchers and clinicians to deliver innovative treatment techniques and assessments that, when combined with conventional therapy, maximize human performance and improve treatment outcomes. A virtual environment is defined as a simulation of a real-world environment and can range from non-immersive to fully immersive. Fully immersive VEs used in clinical assessment and biomechanical research include head-mounted displays, cave automatic virtual environments (CAVE), and Computer Aided Rehabilitation Environments (CAREN).

2.4.1 Head-mounted display

A head-mounted display is a wearable device that projects stereoscopic imagery of an augmented or simulated reality. A head-mounted display is typically a part of a helmet, enclosed visor, or glasses. Head-mounted display systems can be integrated with a motion capture system to track head orientation or line-of-sight enabling user interaction with the virtual environment. Advantages of head-mounted displays over other virtual environments are smaller size, lower cost, and portability. Head-mounted display systems that are designed to completely obstruct the user's field-of-view of the real-world require an accurate representation of all physical real-world objects in the virtual reality or risk visual field and proprioceptive mismatch that are linked with simulator sickness [81]. Head-mounted display systems that do not completely obstruct the user's field-of-view can use augmented reality with a virtual display superimposed on the natural field-of-view.

2.4.2 CAVE

The CAVE is a room-sized system where multiple walls and floor are used as projection screens for the virtual scene. This type of virtual environment has a wide field-of-view for angular viewing without tracking head orientation. Users can wear 3D glasses to simulate depth. CAVE systems can also integrate a motion capture system to track user position and enable interaction with the virtual environment.

2.4.3 CAREN

CAREN systems are sold in one of three standard configurations: base, extended, and high-end [82]. All CAREN configurations consist of a six degree of freedom motion platform integrated with force plates, motion capture system, a projection screen, and control software. The hardware components and virtual scene are integrated with control software (D-Flow), based on a modular programming framework [83]. This allows real-time feedback and additional ways to enable the user to interact with the virtual scenario. The base configuration uses a large flat projection screen, the extended configuration includes a large curved projection screen and a treadmill embedded in the motion platform, and the high-end configuration includes a spherical projection screen and a treadmill embedded in the motion platform.

2.5 Self-paced treadmills

Treadmills are an important tool for both clinicians and researchers and are used to examine uninterrupted ambulation in a controlled environment. Traditionally, biomechanics research has been performed on treadmills operating at a fixed speed. Controlling walking speed reduces outcome measure variability when researching task-oriented biomechanics [84, 85]. However, when walking speed was not controlled, decreased walking speed was often used as a stability enhancing strategy [1, 2]. A treadmill with self-paced mode could elicit more natural gait responses within multi-terrain virtual environments that are not possible with conventional treadmills operating at a fixed speed. The ability to alter gait speed may be especially important for transtibial amputees walking over variable terrain.

Many self-paced treadmill solutions have been developed to automatically adjust speed in real-time to match the user's walking speed [16, 86-91]. Most of these solutions use positional feedback. Ultrasound, resistive tether, or motion capture can detect changes in body position and drive a treadmill speed algorithm that maintains body position relative to a reference point [91, 92]. The self-paced treadmill speed algorithm used in this thesis is an implementation of positional feedback, tracking the body relative to a person's initial standing position using motion capture technology.

Force plate derived center-of-pressure and anterior-posterior ground reaction force can also be used to estimate desired walking speed [87]. However, center-of-mass estimation is still needed to maintain user position relative to a reference point (center of treadmill). Yoon et al. implemented a control algorithm using foot swing velocity to predict desired walking speed within a half step, which was advantageous for simulating the intention to stop walking [86].

Fung et al. [16, 92] successfully used a self-paced treadmill for post-stroke gait training and indicated that self-paced control provided a strong sense of presence in a virtual environment. Hak et al. tested able-bodied [12, 93] and unilateral transtibial amputees [12] to determine if the participants would walk slower on a self-paced treadmill with continuous medial-lateral surface translations. Both able-bodied and transtibial groups did not change their walking speed with medial-lateral translations, but walked with shorter, wider, and faster steps [12, 93]. Most recently, Sloot et al. [91] reported few differences between fixed and self-paced treadmill conditions for level walking in an able-bodied population. Kinematics, kinetics, and stride parameters were comparable between conditions, but variability was greater for self-paced conditions. However, Sloot et al. only examined level treadmill walking for able-bodied participants.

Chapter 3: Evaluation of a motion platform embedded with a dual belt treadmill instrumented with two force plates

This chapter addresses objective 1 (evaluate technical aspects of the VE) by testing operating characteristics of the motion platform and treadmill components of the CAREN-Extended system at The Ottawa Hospital Rehabilitation Centre. Additionally, the effect of platform and treadmill operation on force plate signals w examined. Appendix A is associated with this chapter, presenting the results for an additional force plate test.

The content of this chapter was accepted for publication in the *Journal of Rehabilitation Research and Development*. A portion of this chapter was also presented at the 2013 IEEE International Symposium on Medical Measurements and Applications.

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Sinitski EH, Lemaire ED, Baddour N (2013) Characteristics of a Dual Force Plate System Embedded in a Six Degree of Freedom Motion Platform. IEEE International Symposium on Medical Measurements and Application. Gatineau, Quebec, May.

3.1 Abstract

Motek Medical's CAREN-Extended system is a virtual environment primarily used in physical rehabilitation and biomechanical research. This virtual environment consists of a 180° curved projection screen used to display a virtual scene, a 12-camera motion capture system, and a six degree of freedom platform equipped with a dual-belt treadmill and two force plates. The CAREN system enables clinicians and researchers to systematically manipulate the environment (e.g., walking surface or visual field) to suit the individual needs of a patient or the specific requirements of a research study. This goal of this paper is to investigate the performance characteristics associated with a "treadmill – motion platform" configuration and how system operation can affect the data collected; including, static and dynamic characteristics of the motion platform, treadmill belt speed variability, force plate signals during treadmill and platform operation, and force plate baseline drift. Knowledge of performance characteristics and their effect on outcome data is crucial for effective design of CAREN research protocols and rehabilitation scenarios.

3.2 Introduction

The Computer Aided Rehabilitation Environment (CAREN – Motek Medical, Amsterdam, Netherlands) is a virtual environment and a rehabilitation aid used in research and clinical settings. This system enables innovative rehabilitation techniques and comprehensive evaluation measurements that provide insights into a person's recovery process [17]. In addition to clinical use, researchers employ the CAREN for scientific inquiry, furthering knowledge of walking stability [94], traumatic brain injury [4], and neurorehabilitation [95].

The CAREN-Extended system configuration integrates a motion capture system, six degree of freedom motion platform, instrumented treadmill, and a virtual scene. The CAREN allows researchers to address a number of research questions by enabling manipulation of the standing or walking surface, visual scene, and by providing real-time biofeedback or enabling interaction with the virtual environment.

The CAREN system's discrete technologies are not new, but aggregating these systems for rehabilitation or research is currently state-of-the-art. Understanding this system's performance characteristics and limitations are crucial for effective design of research protocols or formulation of clinical treatment plans. Previously, Lees et al. [96] examined response characteristics of the CAREN's motion platform using sine and ramp input functions, which are commonly used to expose users to balance and postural perturbations. Lees and colleagues provided an amplitude-frequency analysis to understand the best operational range for standing balance protocols [96].

One particular challenge with the CAREN-Extended framework is integrating force plates within the motion platform. In a typical motion analysis laboratory, force plates are secured in a level walkway, shielded from electromagnetic interference, and isolated from environmental vibrations. Conventional treadmill systems instrumented with force plates introduce electrical and mechanical noise from treadmill belt motion and the electric motor [97-99]. In the CAREN-Extended system, force plates are embedded in a treadmill and 6DoF platform. Force plate data is a key component of gait analysis and is typically used as an input of the inverse dynamic model to estimate multiple biomechanical parameters such as torque at the ankle, knee and hip. Preliminary findings demonstrated that platform and treadmill operation affected the ground reaction force (GRF) signals [100], which will affect torque derived from the inverse dynamic model.

The goal of this paper is to evaluate performance characteristics of the motion platform and treadmill, and to examine the effects of different modes of CAREN operation on force measurements. We hypothesized that platform and treadmill operation is consistent with manufacture specifications, and that platform movement and treadmill operation affect force plate measurements. Since these systems are emerging as tools for research and clinical rehabilitation decision-making, a better understanding of technical performance is needed to develop protocols for specific human movement evaluations and to guide data collection and analysis methodologies. The performance tests cover static and dynamic characteristics of the motion platform, treadmill belt speed variability, force plate signals analysis during treadmill and platform operation, ambulation, and baseline drift.

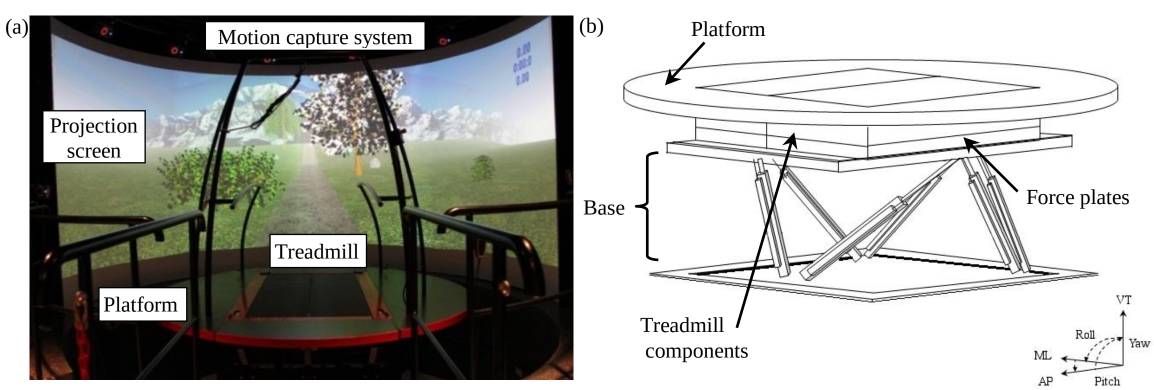

Figure 3.1. CAREN-Extended system (a) and the motion platform embedded with a dual-belt treadmill system and two force plates (b).

3.3 Methods

3.3.1 Equipment description

A commercially available CAREN-Extended system (Figure 3.1) incorporates a 12-camera (MX T20S) Vicon motion capture system (Vicon, Oxford, UK), a 3 meter diameter Sarnicola hydraulic motion platform capable of six degrees of freedom (Sarnicola Simulation Systems, Inc., Conklin, NY), a Bertec 1x2 meter dual-belt treadmill instrumented with two force plates (Bertec Corp., Columbus, OH), a 180° curved projection screen, and four F10 AS3D projectors (projectiondesign, Fredrikstad, Norway). The six hydraulic actuators are connected in a Stewart configuration and controlled independently to enable motion in six degrees of freedom: sway or medial-lateral translation (ML), surge or anterior-posterior translation (AP), heave or vertical translation (VT), pitch, yaw, and roll (Figure 3.1).

3.3.2 Data collection

Three reflective markers were secured to the platform to track platform motion. Vicon Nexus version 1.8.2 software (Vicon, Oxford UK) was used to record marker positions at 100 Hz and force plate data at 1000 Hz. Force plate baselines were reinitialized prior to each performance test or trial. This involved a hardware zeroing of the force plates, while unloaded, stationary, and level. The D-Flow 3.10.0 Platform Module (Motek Medical, Amsterdam, NL) was used to control platform translation and rotation. D-Flow is the control software that integrates hardware components (e.g., treadmill and motion platform) and the virtual scenario using a modular programming framework [83]. The D-Flow platform safety filter parameter was set to 1, which was equivalent to the inverse of the cut-off frequency for a real-time 2nd order Butterworth low pass filter of 1Hz.

Visual3D 4.96.4 (C-Motion Inc., Germantown, MD) and Matlab 2010a (The Mathworks, Matwick, MA) were used to analyze each performance test. Visual3D was used to model the platform as a kinematic object (segment) and to calculate platform translation and rotation.

3.3.2.1 Motion platform testing

The motion platform’s static and dynamic performance characteristics were examined. For the static analysis, the platform’s position was examined throughout the operation range-of-motion in 5 cm or 5° intervals (Table 3.1) for four trials per condition. The mean platform angle or position over 1 s was determined for each trial, when the platform was stationary, and compared to the input position or rotation.

To examine the dynamic performance, the platform was translated or rotated over progressively larger distances. At the start of each trial, the platform was positioned at the laboratory origin and then translated or rotated. After a few seconds, the platform was translated or rotated back to the starting position. This was repeated twice for each of the conditions listed in Table 3.2 and for three D-Flow safety filter input settings (1, 2.5, and 5), where larger safety filter values resulted in more gradual platform movement.

Table 3.1. Motion platform static testing range.

Translation (cm)	ML and AP	-25 to 25 in 5 cm intervals
	VT	-20 to 32 in 5 cm intervals
Rotation (deg)	Pitch and Roll	-20 to 20 in 5° intervals
	Yaw	-20 to 20 in 5° intervals

Table 3.2. Motion platform dynamic testing ranges.

Translation (cm)	ML and AP	5, 10, 15, 20, 25
	VT	5, 10, 15, 20, 25, 30
Rotation (deg)	Pitch and Roll	5, 10, 15, 20
	Yaw	5, 10, 15

Platform velocity, acceleration, and deceleration were calculated for translation trials; and platform angular velocity, angular acceleration, and angular deceleration were calculated for rotation trials (Figure 3.2). Total platform transition time was defined from the time the platform velocity was greater than 0 m/s (Figure 3.2; T0) to the time when the platform velocity was 0 m/s and thus stationary (Figure 3.2; T2). Peak platform transition time was defined as the time from initial to peak platform position (Figure 3.2; T1). At the start of each trial, the force plates were reinitialized while the platform was level and stationary. During the trials, the platform changed orientation and the forces due gravity and platform weight were represented in the force signal (Figure 3.3a). To correct for these orientation-based forces, an offset (Figure 3.3b) was subtracted from the original force data so that the force data was zero when the platform was stationary (Figure 3.3c). Mean and two standard deviations of the resulting force signal were calculated to quantify the force due to platform acceleration (Figure 3.2).

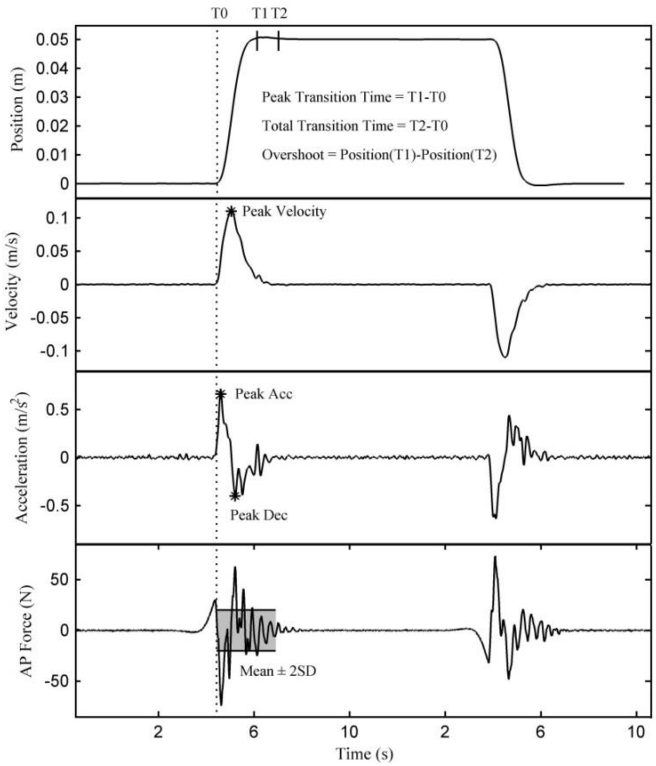

Figure 3.2. Parameters for motion platform testing.

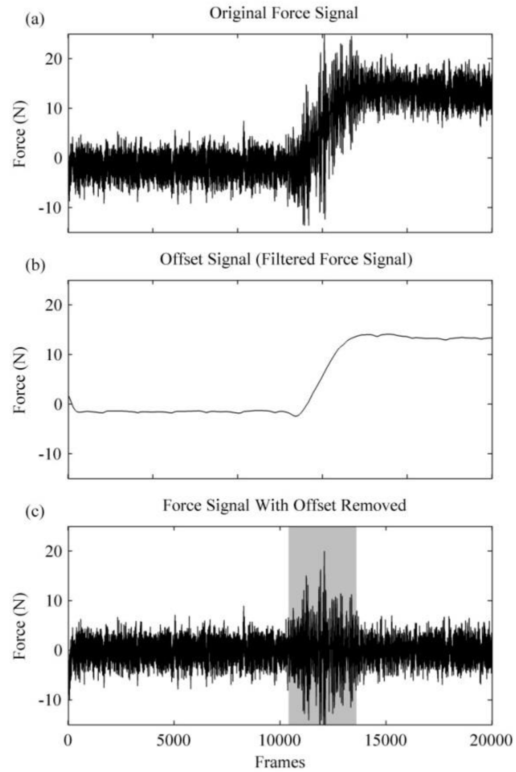

Figure 3.3. Force plate output for level platform transitioning to an incline orientation. (a) is the raw force signal, (b) is the filtered signal representing the offset (1Hz, low pass), and (c) is the raw signal minus offset. The shaded area represents the period when the platform was moving.

3.3.2.2 Force plate testing

Force plate measurements were examined when the treadmill was in operation, during ambulation, and over time to observe the baseline drift. Data from two force plates were recorded as each treadmill belt speed increased from 0–5 m/s, in 0.5 m/s increments. Mean and two standard deviations over 5 s of force data were calculated for each speed. The power spectral density (Welch method [101]) of the force plate signal was estimated for the ML, AP, and VT directions and each treadmill belt speed.

Force plate measurements were also examined while a person walked at a self-selected pace (1.1 m/s) on one force plate to determine the effect of foot strike impact forces on the other (unloaded) force plate. Mean and two standard deviations of force data over five foot strikes were calculated to examine the effect of foot contact on the unloaded force plate. The power spectral density of the unloaded force plate signal was also examined for the ML, AP, and VT directions.

Finally, force plate measurement drift was examined during an unloaded condition and a walking condition. All data were collected after following the standardized power-up sequence and the force plate amplifiers reached optimal operational condition. For each condition, the force plates were reinitialized and a 5 s trial was recorded to capture the baseline signal. For the unloaded condition, a 5 s trial was recorded every five minutes for a 40 minute period, without a person walking on the platform. For the walking condition, a person walked on the treadmill for five trials where each trial was five minutes in length. Five seconds of force data were recorded between each trial, when the participant was off the platform. The mean forces between trials were examined.

3.3.2.3 Treadmill testing

Treadmill speed was examined with and without a person walking on the treadmill to determine the difference between input settings and measured treadmill belt speed. Three reflective markers were attached to the spokes of a distance-measuring wheel, which consisted of a single wheel attached to a handle that can be pushed or pulled alongside a person. To examine treadmill speed without a person walking on the treadmill, a person stood on the platform next to the treadmill, holding the distance-measuring wheel on the treadmill belt. To examine speed with a person walking on the treadmill, a person walked on one belt while holding the distance wheel out in front of their feet on the same treadmill belt. Markers were also attached to the person’s feet to calculate treadmill speed from the foot marker trajectories during single limb stance. Treadmill belt speed was increased from 0.5 – 3.0 m/s in 0.5 m/s increments without a person on the treadmill, and from 0.5 – 2.0 m/s in 0.25 m/s increments with a person walking on the treadmill. Each increment was approximately 10 s in duration. Motion data were collected for left and right treadmill belts separately while the platform was level, 7° incline (pitch), and 7° decline (pitch).

The distance wheel was modeled using Visual3D, where the origin was at the center of the distance wheel. Angular velocity of the distance-measuring wheel was calculated and converted to linear velocity of the treadmill belt. Mean and standard deviation were calculated over 3 s for each treadmill belt and each speed increment, without a person on the treadmill. Treadmill speed was also calculated from foot marker AP velocity to evaluate the use of foot marker data as a method to determine treadmill speed. Five consecutive frames of foot marker velocity were selected during single leg stance. These data corresponded to the smoothest part of the velocity curve when the foot was flat on the treadmill. Treadmill velocity for the same period was also derived using the distance wheel. Mean velocity over 10 steps was calculated for both methods to evaluate foot marker velocity as a measure of treadmill belt speed.

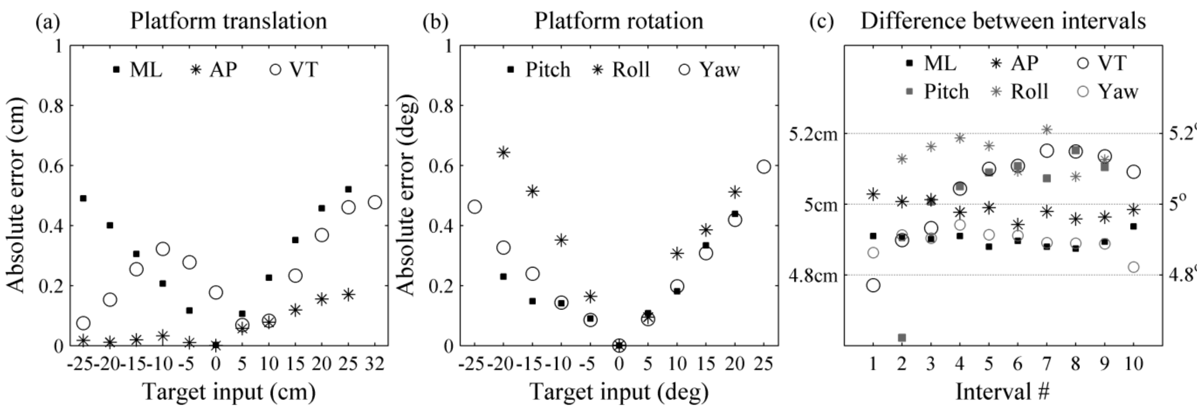

Figure 3.4. Difference between D-Flow input and measured platform position for platform translation (a) and rotation (b) operational range, for one trial. Difference between each platform transition (c), where 5cm or 5° was the desired transition.

3.4 Results

3.4.1 Motion platform testing

The static performance for platform translation and rotation was consistent with the input specified in D-Flow. Differences between the input and measured platform orientations were on average less than 0.5 cm for translation trials (Figure 3.4a) and less than 1° for rotation trials (Figure 3.4b), where larger errors were observed at the end of the test range. Additionally, platform movement in the in the other axes (e.g., AP and VT axes for ML translation trials) were on average less than 0.5 cm for translation trials and less than 0.5° for rotation trials.

Maximum platform velocity, acceleration, and deceleration increased as the distance the platform travelled increased (Table 3.3 and Table 3.4). Maximum deceleration was smaller than maximum acceleration for the same distance travelled. Platform peak transition time was approximately 1 s for a default D-Flow safety filter, set to 1, and increased to approximately 4 s for the largest safety filter setting (Table 3.3 and Table 3.4). Peak platform orientation was often greater than the specified input (overshoot) before gradually moving to the final orientation. As the D-Flow safety filter increased, the platform total transition time and overshoot increased, and the platform acceleration decreased (Table 3.3 and Table 3.4). When the platform was stationary, force signals were less than 3 N and force signals increased when the platform was moving. Force signals increased as platform acceleration increased and were largest in the direction of platform motion (Figure 3.5). Even the most gradual platform movement (safety filter = 5) introduced force plate noise when compared to a stationary platform, but the noise was less than 30 N (Figure 3.5).

Table 3.3. Dynamic performance characteristic for platform translation including platform velocity, acceleration, deceleration, peak transition time, total transition time, and overshoot.

	Peak Velocity (m/s)			Peak Acceleration (m/s^2)			Peak Deceleration (m/s^2)			Peak Transition Time (s)			Total Transition Time (s)			Overshoot (m)		
	ML	AP	VT	ML	AP	VT	ML	AP	VT	ML	AP	VT	ML	AP	VT	ML	AP	VT
D-Flow safety filter = 1																		
0-5 cm	0.11	0.11	0.12	0.52	0.63	0.66	0.32	0.38	0.44	1.2	1.1	1.1	1.2	1.1	1.1	0.000	0.000	0.000
0-10 cm	0.24	0.25	0.25	1.23	1.24	1.30	0.76	0.78	0.86	1.1	1.0	1.0	1.4	1.2	1.5	0.000	0.000	0.001
0-15 cm	0.37	0.38	0.37	1.96	1.91	1.64	1.20	1.39	1.01	1.1	1.0	1.0	1.5	1.3	1.6	0.001	0.001	0.001
0-20 cm	0.50	0.52	0.49	2.79	2.68	2.29	1.90	1.86	1.43	1.0	1.0	1.0	1.5	1.6	1.6	0.002	0.002	0.002
0-25 cm	0.64	0.65	0.59	3.32	3.17	2.44	2.50	2.48	2.41	0.9	0.9	1.0	1.5	1.5	1.6	0.003	0.003	0.004
D-Flow safety filter = 2.5																		
0-5 cm	0.05	0.05	0.06	0.17	0.23	0.38	0.10	0.20	0.38	2.1	2.1	2.1	3.5	3.1	3.1	0.001	0.001	0.002
0-10 cm	0.11	0.11	0.11	0.30	0.43	0.42	0.14	0.22	0.39	2.1	2.0	2.0	3.4	3.3	3.5	0.003	0.003	0.003
0-15 cm	0.16	0.17	0.17	0.45	0.56	0.62	0.21	0.29	0.36	2.0	2.0	2.0	3.5	3.3	3.5	0.005	0.004	0.005
0-20 cm	0.22	0.23	0.23	0.62	0.80	0.75	0.30	0.33	0.43	2.1	2.0	2.0	3.5	3.4	3.3	0.007	0.006	0.007
0-25 cm	0.28	0.29	0.29	0.82	0.96	0.95	0.35	0.41	0.41	2.0	2.0	2.0	3.6	3.5	3.3	0.009	0.009	0.009
D-Flow safety filter = 5																		
0-5 cm	0.03	0.03	0.03	0.10	0.18	0.34	0.09	0.16	0.25	3.9	3.7	3.5	6.0	5.9	5.8	0.002	0.002	0.002
0-10 cm	0.05	0.06	0.06	0.13	0.18	0.33	0.09	0.13	0.25	3.7	3.7	3.5	6.2	6.4	6.0	0.004	0.004	0.004
0-15 cm	0.08	0.09	0.09	0.18	0.24	0.40	0.10	0.11	0.16	3.8	3.7	3.8	6.1	6.1	6.2	0.005	0.005	0.006
0-20 cm	0.11	0.11	0.12	0.20	0.29	0.38	0.09	0.14	0.13	3.7	3.7	3.6	6.4	6.3	6.0	0.008	0.008	0.008
0-25 cm	0.14	0.14	0.15	0.28	0.42	0.37	0.12	0.15	0.15	3.7	3.7	3.6	6.9	6.2	6.2	0.010	0.010	0.010

Table 3.4. Dynamic performance characteristic for platform rotation including platform velocity, acceleration, deceleration, peak transition time, total transition time, and overshoot. For the yaw condition, 0-20 deg test was not performed.

	Peak Velocity (deg/s)			Peak Acceleration (deg/s^2)			Peak Deceleration (deg/s^2)			Peak Transition Time (s)			Total Transition Time (s)			Overshoot (deg)		
	Pitch	Roll	Yaw	Pitch	Roll	Yaw	Pitch	Roll	Yaw	Pitch	Roll	Yaw	Pitch	Roll	Yaw	Pitch	Roll	Yaw
D-Flow safety filter = 1																		
0-5 deg	12.0	12.1	11.3	57.8	66.8	55.7	41.9	40.9	31.5	1.1	1.1	1.1	1.1	1.3	1.1	0.0	0.0	0.0
0-10 deg	25.4	26.0	24.4	124.4	139.9	126.7	87.4	103.3	80.1	1.0	1.0	1.1	1.6	1.0	1.6	0.1	0.0	0.1
0-15 deg	38.6	39.3	36.5	194.3	210.4	193.3	135.5	154.4	125.7	1.0	1.0	1.0	1.6	1.5	1.6	0.2	0.1	0.2
0-20 deg	50.9	52.2	na	236.6	259.3	na	194.9	207.4	na	1.0	1.0	na	1.5	1.6	na	0.2	0.3	na
D-Flow safety filter = 2.5																		
0-5 deg	5.5	5.5	5.2	25.5	40.2	16.2	16.4	27.8	12.9	2.0	2.1	2.1	3.4	3.4	3.3	0.1	0.2	0.2
0-10 deg	11.2	11.6	10.8	32.6	37.3	36.3	19.6	18.9	15.7	2.1	2.0	2.1	3.4	3.4	3.3	0.3	0.3	0.3
0-15 deg	17.3	17.6	16.6	51.9	50.3	54.5	25.5	24.8	19.3	2.0	1.9	2.1	3.6	3.5	3.3	0.5	0.5	0.5
0-20 deg	23.0	23.4	na	66.6	79.5	na	33.7	45.9	na	2.0	1.9	na	3.5	3.3	na	0.7	0.6	na
D-Flow safety filter = 5																		
0-5 deg	3.0	3.1	2.8	15.0	19.1	11.0	9.1	21.7	6.0	3.7	3.4	3.6	6.2	5.6	6.3	0.2	0.2	0.2
0-10 deg	5.8	6.0	5.6	20.5	35.0	13.8	17.1	27.7	9.6	3.8	3.6	3.8	6.3	5.7	6.2	0.4	0.4	0.4
0-15 deg	8.8	9.0	8.4	23.4	33.1	19.5	16.9	32.4	11.6	3.8	3.7	3.8	6.2	5.7	6.1	0.6	0.5	0.6
0-20 deg	11.7	12.0	na	31.5	32.4	na	16.1	31.3	na	3.8	3.6	na	6.1	6.2	na	0.8	0.7	na

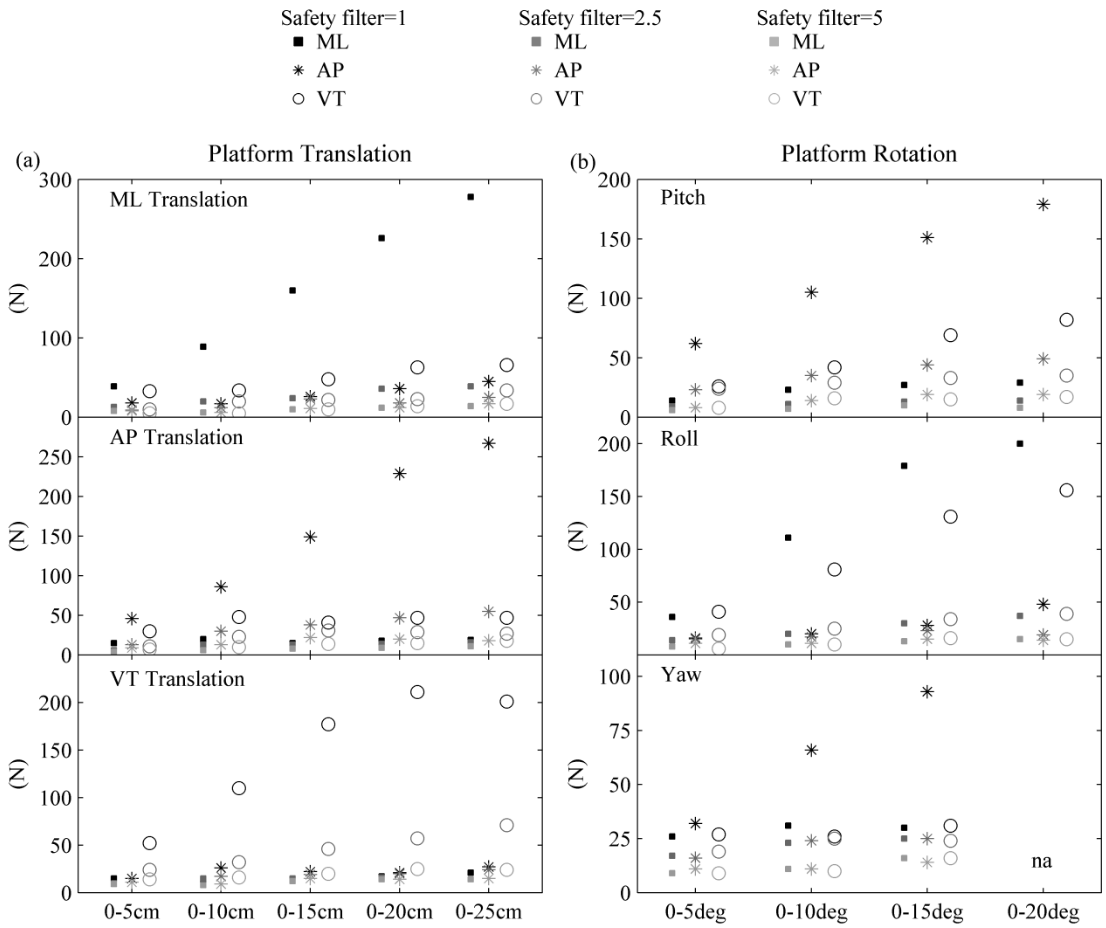

Figure 3.5. Two standard deviations of force measurements due to platform acceleration for progressively larger platform movement.

3.4.2 Force plate testing

Force plate noise was less than 3 N when the treadmill belts were stationary (0 m/s) and increased with increasing treadmill belt speed. This noise would be predominantly electrical when treadmill belts were stationary and a combination of electrical and mechanical during treadmill operation. Force plate noise was less than 10 N for 0.5–2.0 m/s treadmill speeds and was between 10 and 40 N for speeds of 2.5–5.0 m/s. The power spectral density showed that most signal noise due to the treadmill components occurred at frequencies higher than 20 Hz (Figure 3.6a) and could be effectively removed using a low pass filter with a 20 Hz cut-off frequency (Figure 3.6b).

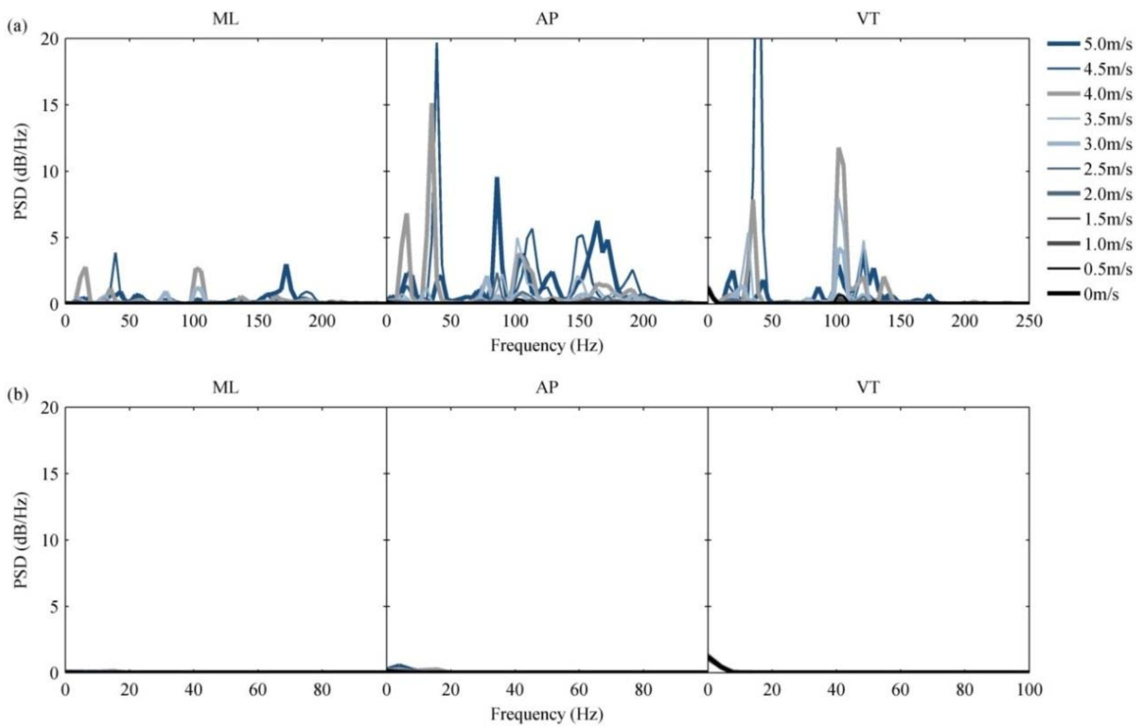

Figure 3.6. Power spectral density of the unfiltered force signal (a) and filtered force signal (b) for each axis (ML=medial-lateral, AP=anterior-posterior, and VT=vertical axes) as treadmill belt speed increased from 0 – 5m/s. Force data were filtered using a 4th order Butterworth low pass filter with a cut-off frequency of 20 Hz (b).

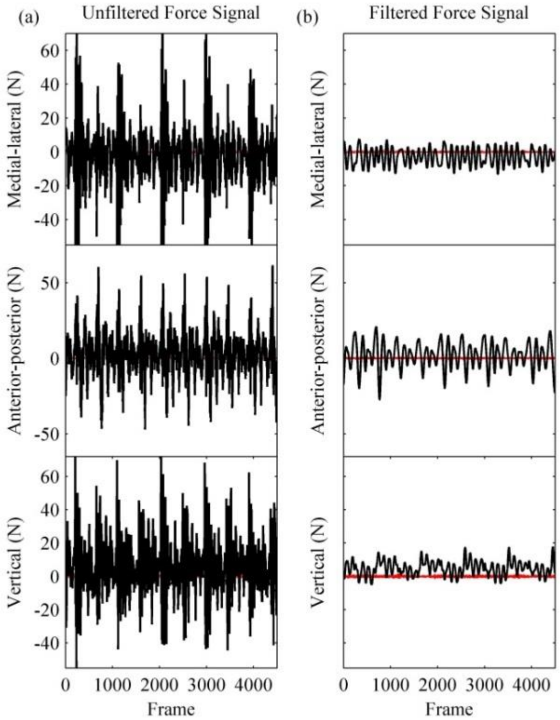

Figure 3.8. Unfiltered force signal artifact (a) and filtered force signal artifact (b) from the unloaded force plate while a person only walked on the other force plate. The force measurements on the unloaded force plate should be 0 N; however a force signal artifact was observed in the force signal. Each peak in this force signal (a) represents a foot strike on the other force plate. Force data were filtered using a 4th order Butterworth low pass filter with a cut-off frequency of 20 Hz (b). The red line at 0 N on all graphs represents the force signal without a person walking on the platform.

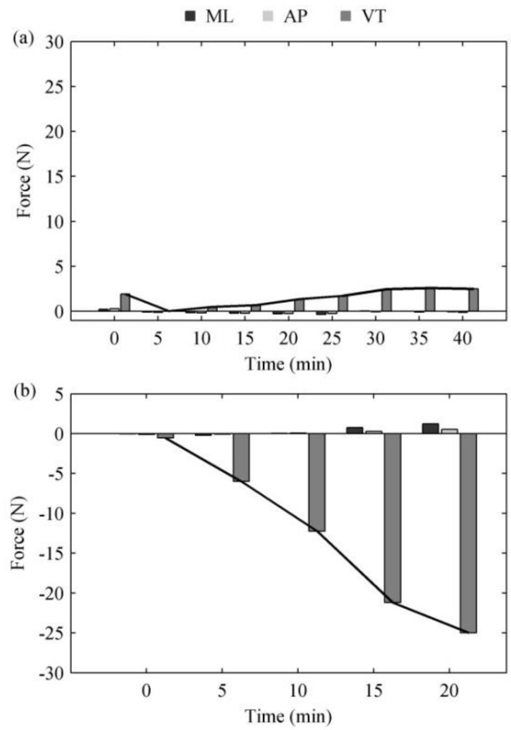

Figure 3.7. Force signal baseline drift for each axis (ML=medial-lateral, AP= anterior-posterior, and VT=vertical axes) without a person walking on the treadmill (unloaded condition) (a) and with a person walking on the treadmill (walking condition) (b).

When a person walked on the platform, the platform tracking markers exhibited more vertical translation than vertical translation observed without a person walking on the platform, but this difference was less than 1mm. With a person walking on one force plate, the other or unloaded force plate measurements should be 0 N. However, a force signal artifact was observed on the unloaded force plate equivalent to ± 40 N, averaged over all force channels (Figure 3.8a). The power spectral density of this force signal artifact resulted in frequency components between 0–40 Hz, which overlaps the frequency components of ground reaction forces [102] and cannot be easily removing by filtering techniques. Applying a low pass filter was only able to reduce the force signal artifact to less than 20 N (Figure 3.8b).

Without a person walking on the platform, ground reaction forces drifted less than 3 N over a 40 minute period (Figure 3.7a). When a person walked on a level platform, the AP and ML ground reaction forces drifted less than 5 N over a 25 minute period, but VT drifted approximately 5 N every five minutes (Figure 3.7b).

Table 3.5. Mean and standard deviation of treadmill belt speed for one belt with (walking condition) and without (unloaded condition) a person walking on the treadmill. For the walking condition, treadmill belt speed was calculated from foot marker data and distance wheel data for 10 steps.

Input speed (m/s)	Platform Condition			
	Level	Decline	Incline	% Error Mean
Unloaded condition – calculated from distance wheel				
0.50	0.52 (0.01)	0.52 (0.01)	0.52 (0.01)	4.7
1.00	1.03 (0.01)	1.04 (0.01)	1.04 (0.01)	3.7
1.50	1.55 (0.01)	1.55 (0.01)	1.55 (0.01)	3.3
2.00	2.06 (0.01)	2.07 (0.01)	2.07 (0.01)	3.3
2.50	2.58 (0.01)	2.58 (0.01)	2.58 (0.01)	3.2
3.00	3.10 (0.02)	3.10 (0.01)	3.10 (0.01)	3.3
Average				*3.6*
Walking condition – calculated from distance wheel				
0.50	0.51 (0.06)	0.53 (0.05)	0.51 (0.03)	3.3
0.75	0.75 (0.05)	0.78 (0.03)	0.74 (0.04)	1.8
1.00	1.03 (0.03)	1.04 (0.05)	1.00 (0.04)	2.3
1.25	1.29 (0.05)	1.31 (0.05)	1.27 (0.03)	3.2
1.50	1.54 (0.07)	1.58 (0.06)	1.52 (0.04)	3.1
1.75	1.80 (0.05)	1.86 (0.07)	1.75 (0.04)	4.0
2.00	2.12 (0.05)	2.09 (0.06)	2.07 (0.06)	4.7
Average				*3.2*
Walking condition – calculated from foot marker				
0.50	0.52 (0.00)	0.52 (0.01)	0.52 (0.00)	4.0
0.75	0.79 (0.00)	0.76 (0.01)	0.79 (0.01)	4.0
1.00	1.05 (0.01)	1.04 (0.01)	1.05 (0.01)	4.7
1.25	1.31 (0.01)	1.30 (0.01)	1.31 (0.00)	4.5
1.50	1.57 (0.01)	1.56 (0.02)	1.58 (0.01)	4.7
1.75	1.84 (0.01)	1.81 (0.02)	1.82 (0.01)	4.2
2.00	2.11 (0.01)	2.10 (0.01)	2.11 (0.01)	5.3
Average				*4.5*

3.4.3 Treadmill testing

Treadmill speed was consistent across the platform conditions tested (level, incline, and decline) with and without a person walking on the treadmill (Table 3.5). Treadmill speed variability was greater when a person walked on the treadmill and the variability was greater at faster speeds. The measured treadmill speed was approximately 3% faster than the input speed without a person walking on the treadmill, and approximately 4% faster when a person walked on the treadmill. Treadmill speed calculated from foot marker velocity during single leg stance was consistent, with a standard deviation less than 0.02 m/s across 10 steps. The difference between treadmill belt speed calculated from foot marker velocity data and treadmill belt speed calculated from the distance wheel was less than 2%.

3.5 Discussion

This paper examined a number of operational parameters and their effects on force plate data collected in a motion analysis system equipped with a six degree of freedom platform. Understanding how modes of operation affect force plate signals and how to minimize possible interferences is important for clinicians and researchers when developing rehabilitation treatment plans or research protocols.

Although differences between input and measured platform orientation were small, the difference between each increment was consistently slightly smaller or larger (Figure 3.4c), instead of the intended increment of 5cm or 5°. As a result, this error was compounded over the total platform movement and was only reset when the platform returned to a neutral or settled position. Including platform tracking markers for CAREN data collection sessions will allow the user to determine if errors in platform motion have occurred and determine force plate corner locations for slope conditions.

The results from this research also showed how various equipment settings affect platform acceleration. Platform transition time was constant over different platform travel distances (Table 3.3 and Table 3.4), for a given safety filter setting, and the platform acceleration varied. Increasing platform travel distance resulted in larger platform accelerations, corresponding to larger forces applied to a person (Figure 3.5). Both platform acceleration and safety filter settings affected the magnitude of this force. While larger forces might be useful in creating more challenging environments, when investigating postural balance or walking stability, an understanding of these effects are important when attempting to control for perturbing force. Furthermore, there is currently no accepted method to account for platform acceleration effects on force data; therefore, kinetic analyses should only process gait cycles when the platform is stationary.

Depending on the scenario, gradual platform transition or smaller perturbing force may be preferable to ensure patient safety for people with mobility disabilities. The D-Flow Platform Module safety filter parameter can be used to control platform acceleration; however, this setting affects all platform movement in an application. Furthermore, an application may require varied platform travel distance during a session or less perturbing force than the Platform Module safety filter parameter allows. To address this problem, the D-Flow Filter module can be used to generate custom and activity-specific platform transitions [100]. Generally, a safety filter value of 5 (0.2 Hz low pass filter) produced a comfortable, 3 s platform peak transition time for a 7° slope. For people with mobility disabilities, slower transition speeds may be required for transitioning from a level platform to steeper slopes.

An overshoot was observed for platform conditions tested, and increased as D-Flow safety filter increased or platform travel distance increased (Table 3.3 and Table 3.4). As a result, larger overshoots resulted in longer total platform transition times. Although, the platform moved very slowly during the final transition period (approximately 5% of peak velocity), this platform motion still resulted in force plate signal noise. These findings demonstrated the importance of selecting gait cycles after the platform reached its final position for kinematic and kinetic analysis.

Consistent with the hypothesis, treadmill operation affected force plate signals. In this study, the signal noise due to the treadmill operation was small and negligible for speeds less than 2.0 m/s. Similar to data reported by Paolini [97], this noise can be effectively removed by standard low pass filtering methods for treadmill systems.

Force and moment signals did not drift appreciably when the force plates were unloaded. However, there was a consistent vertical drift of 5 N every five minutes when the force plates were in continuous use (Figure 3.7b). For a level platform, vertical axis baseline drift will affect vertical ground reaction force magnitudes and, in turn, automatic gait event identification, custom event triggers in D-Flow, and outcome measures from inverse dynamics (e.g., sagittal joint forces, moments, and powers). However, we only examined force plate baseline drift when the platform was level, where vertical ground reaction force is equal to the force plate’s vertical force measurement. For a non-level platform orientation (e.g., sloped), baseline drift could exist in other force plate axes since the vertical ground reaction force is a resultant of the three force plate axes. Since larger baseline drift can be expected for long data collections, force plates should be zeroed frequently when force data is recorded or utilized in the D-Flow scenario. Depending on the type of CAREN session, it may not be practical for the participant to get on and off the platform to reinitialize the force plate baseline. In this case, post-processing techniques can be used to account for the drift by using force data during the swing phase of gait [97].

Additionally, impact forces during walking also affected force plate signals on the opposite force plate. A force signal artifact was observed on the unloaded force plate while a person walked on the opposite force plate, and this artifact could not be easily removed with standard filtering techniques. This artifact may reflect platform vibration, slight structural deformations of the frame, or response characteristics of the hydraulics resulting from the impact force of foot strikes. Furthermore, since both force plates are secured to the same base, they are not isolated from each other, and any vibrations through the structure will affect measurements on both force plates. Power spectral density analysis showed that the signal artifact frequency components overlapped with the gait frequency components. Although we examined the force signal artifact for one walking velocity, the artifact's magnitude should increase when impact forces are larger due to mass, faster walking velocities, decline walking, or pathological gait (e.g., lower limb amputation). Filtering may be inadequate for completely removing these artifacts and larger artifacts may appear as a "transient force" in ground reaction force signals. Future research with multiple walking speeds and participant demographics could define the scope of signal artifacts due to gait on this six degree of freedom motion platform.

Treadmill belt speed was consistently 3% faster than the specified input speed for the treadmill system used in this paper. Treadmill belt speed variability was larger when a person walked on the treadmill and was consistent with previous research that demonstrated treadmill belt speed variability is also influenced by the person's mass [103, 104]. Using foot markers to determine treadmill speed was similar to the treadmill speed measured using the distance wheel. This is a useful method for calculating temporal-spatial parameters such as stride length and step length, particularly when using a treadmill system with a self-paced (variable speed) feature.

3.6 Conclusions

This research highlighted a number of considerations when operating and interpreting the data collected using the CAREN-Extended system. Specifically, factors that may affect force measurements collected from force plates embedded in a motion platform were examined. These factors have to be considered to ensure data validity and to avoid misinterpretation of the results. Motion and force signal artifact were observed even when the platform was stationary. While platform acceleration affects force signals, there is no accepted method to account for these effects. Future research is necessary to account for platform acceleration and continuous platform movement in order to obtain accurate force data for the new generation of virtual rehabilitation systems. Since CAREN system hardware can vary between installations, these tests should be repeated at individual sites.

3.7 Acknowledgements

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Chapter 4:Fixed and self-paced treadmill walking for able-bodied and transtibial amputees in a multi-terrain virtual environment

This chapter addresses objective 2 (evaluate self-paced treadmill mode) by examining gait biomechanics when walking on a self-paced treadmill in order to understand how people vary walking speed and to understand limitations when using this mode.

The following appendices are associated with this chapter:

1. Appendix B includes a sample video of the virtual scenario.
2. Appendix C includes Visual3D pipeline used to subtract baseline drift and gravity effects from the force plate measurements.
3. Appendix D includes Matlab code for determining treadmill belt velocity from foot markers.
4. Appendix E includes the results tables associated with this manuscript.
5. Appendix F-Appendix H include the information sheet and consent form for this research protocol.

A portion of this chapter was presented at the 3rd Annual Canadian Institute for Military & Veteran Health Research Forum (CIMVHR) in Kingston, Ontario, November 2012.

Sinitski EH, Lemaire ED, Baddour N, Besemann M, Dudek NL (2012) The Effect of Self-paced Treadmill Walking Conditions in a Virtual Environment. Canadian Institute for Military & Veteran Health Research Forum (CIMVHR). Kingston, ON.

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Sinitski EH, Lemaire ED, Baddour N, Besemann M, Dudek NL, Hebert JS (2014) Fixed and Self-paced Treadmill Walking for Able-bodied and Transtibial Amputees in a Multi-terrain Virtual Environment.

4.1 Abstract

A self-paced treadmill automatically adjusts speed in real-time to match the user's walking speed, potentially enabling more natural gait than fixed-speed treadmills. This research examined walking speed changes for able-bodied and transtibial amputee populations on a self-paced treadmill in a multi-terrain virtual environment and examined gait differences between fixed and self-paced treadmill speed conditions.

Twelve able-bodied (AB) individuals and twelve individuals with unilateral transtibial amputation (TT) walked in a park-like virtual environment with level, slopes, and simulated uneven terrain scenarios. Temporal-spatial, range-of-motion, and peak ground reaction force parameters were analyzed.

For self-paced, all participants significantly varied their walking speed ($p<0.001$) and reduced speed for uphill and hilly activities ($p<0.001$). TT also reduced speed downhill ($p<0.001$). Generally, differences between fixed and self-paced speed conditions for temporal-spatial, range-of-motion, and ground reaction force parameters were no longer significantly different with a speed covariate. However, for uphill walking, both groups decreased stride length during self-paced trials, and increased stride length during fixed-speed trials to maintain the constant speed ($p<0.01$).

The results from this study demonstrated self-paced treadmill mode is important for virtual reality systems with multiple movement scenarios in order to elicit more natural gait across various terrain. Fixed-speed treadmills may induce gait compensations to maintain the fixed speed.

4.2 Introduction

Treadmills are important for both clinicians and researchers when examining uninterrupted ambulation in a controlled environment. Traditionally, biomechanics research has been performed on treadmills operating at a fixed speed. Controlling walking speed reduces outcome measure variability when researching task-oriented biomechanics [84, 85]. However, when walking speed was not controlled, decreased walking speed was often used as a stability enhancing strategy [1, 2]. A treadmill with self-paced mode could elicit more natural gait responses within multi-terrain virtual environments that are not possible with conventional treadmills operating at a fixed speed. The ability to alter gait speed may be especially important for transtibial amputees walking over variable terrain.

Many self-paced treadmill solutions have been developed to automatically adjust speed in real-time to match the user's walking speed [16, 87, 90, 91]. Positional feedback is commonly used for self-paced speed control. Sensors or motion capture systems detect body position and drive a treadmill speed algorithm that maintains body position relative to a reference point [91, 92]. Fung et al. [16, 92] successfully used a self-paced treadmill for post-stroke gait training and suggested that self-paced control provided a strong sense of presence in a virtual environment. Sloot et al. [91] reported that average kinematics and stride parameters were comparable between fixed and self-paced treadmill conditions, but variability was greater for the self-paced condition. However, Sloot et al. only examined level treadmill walking for able-bodied participants. Hak et al. tested able-bodied [12, 93] and unilateral transtibial amputees [12] to determine if the participants would walk slower on a self-paced treadmill with continuous medial-lateral surface translations. Both able-bodied and transtibial groups did not change their walking speed with medial-lateral translations, but walked with shorter, wider, and faster steps [12, 93]. Other studies on overground walking reported slower walking speeds with shorter, wider and slower steps for able-bodied and transtibial amputees when walking over other challenging walking surfaces (e.g., loose rocks, wooden blocks, slopes) compared to level ground [32-34, 36, 59].

The purpose of this study was to compare gait biomechanics between fixed and self-paced treadmill speeds for able-bodied (AB) and unilateral transtibial (TT) populations to understand how people vary their walking speed in a stability-challenging virtual environment. For self-paced treadmill walking, we hypothesized that AB and TT participants would reduce walking speed for slopes and simulated uneven terrain, accompanied by changes in gait parameters. Between the fixed and self-paced conditions, differences were expected for temporal–spatial, range-of-motion, and ground reaction force parameters over slopes and simulated uneven terrain for both able-bodied and transtibial amputee participants. Since transtibial amputee and able-bodied gait differs in the real world, these groups may respond differently to self-paced walking in a multi-terrain virtual environment. This research is important for guiding methodological decisions related to treadmill modes when walking in new virtual environments.

4.3 Methods

4.3.1 Subjects

A convenience sample of twelve able-bodied individuals (AB; 11 males, 1 female; mass=80.5±11.8 kg; height=1.78±0.07 m; age=39±11 years) and twelve individuals with unilateral transtibial amputation (TT; 11 males, 1 female; mass=79.9±21.1 kg; height=1.78±0.07 m; age=42±10 years; Table 4.1) were recruited through The Ottawa Hospital Rehabilitation Centre, Canadian Forces Health Services, Glenrose Rehabilitation Hospital, and the community. TT participants were a K-level of three (community ambulator with the ability to traverse most environmental barriers) or four (exceeds basic ambulation skills, exhibiting high impact, stress, or energy levels) and wore their prosthesis daily. The AB group was matched to the TT group based on gender, age, height, and weight. AB limbs were classified as dominant or non-dominant, where the dominant limb was defined as the preferred limb used to kick a ball. All participants provided written informed consent and the study was approved by the ethics review board at each testing institution.

Table 4.1. Transtibial amputee participant demographics.

ID	Gender	Age (yr)	Weight (kg)	Height (m)	Prosthetic side (R/L)	Etiology	Time since amputation (yr)	Level	Prosthetic Foot
S01	M	63	63	1.83	L	Trauma	8	K4	Trias
S02	M	36	96.4	1.80	R	Trauma	5	K4	C-walk
S03	M	45	98.2	1.81	R	Trauma	8	K4	Re-Flex rotate
S04	M	35	67.3	1.73	R	Trauma	1	K4	Variflex
S05	M	44	108	1.86	L	Trauma	1	K3	Variflex
S06	M	50	109	1.85	R	Trauma	1	K4	Triton
S07	M	56	87.3	1.80	L	Trauma	22	K3	Ceterus
S08	M	45	47.3	1.73	R	Trauma	15	K4	Flex foot
S09	M	43	75.9	1.75	R	Osteosarcoma	0.67	K4	Re-Flex rotate
S10	M	37	81.8	1.80	R	Osteosarcoma	3	K4	Variflex
S11	M	29	76.4	1.78	L	Congenital	28	K4	Variflex
S12	F	27	47.7	1.60	R	Other	1	K3	Variflex

4.3.2 Equipment

The CAREN-Extended virtual environment (Motek Medical, Amsterdam, NL; Figure 3.1a, page 33) was used for this study consisting of D-Flow control software, motion capture system (Vicon, Oxford, UK), 180° curved projection screen, and a six degree of freedom motion platform (no handrails) with a 1x2 meter instrumented treadmill (Bertec Corp., Columbus, OH). The

12-camera Vicon system tracked full body kinematics using a six degree of freedom marker set [105] and platform motion using 3 markers, recording at 100 Hz. Anatomical landmarks were defined with a C-Motion digitizing pointer (C-Motion Inc., Germantown, MD). Force plate data were recorded at 1000 Hz and only analyzed when the motion platform was stationary (e.g., level and slopes).

4.3.3 Virtual environment

The test environment consisted of a simulated pathway through a park-like virtual scene, constructed from 20m sections, where each section corresponded to one of eight walking activities (Table 4.2). Participants were asked to complete three trials at a fixed-speed and three trials at a self-paced speed. Each trial consisted of all eight walking activities. All trials started from standing, followed by a period of level walking, and then each subsequent walking activity was separated by a level section. All trials ended with a level section. The total walking distance was 320m for fixed-speed trials and 340m for self-paced speed trials (self-paced trials included a longer initial level walking period). Each trial took approximately 3-5 minutes to complete (see Appendix B for a sample video). Activity and speed condition order were randomized. Some amputee participants were only able to complete two trials for each speed condition due to fatigue.

The self-paced treadmill speed algorithm, as described by Sloot et al. [91] (Methods 2c), incorporated anterior-posterior pelvis position, velocity, and acceleration, referenced to the person's initial standing position (i.e., heels placed at the treadmill's anterior-posterior midline). D-Flow self-paced sensitivity scaling was 0.7 and maximum treadmill acceleration was $1 m/s^2$.

Participants were given an opportunity to acclimate to the virtual scenario at both fixed and self-paced conditions, and all activities (10-20 min). For the fixed-speed condition, a starting speed was calculated according to $\mathrm{Speed} = 0.4\sqrt{\mathrm{leglength} * 9.81}$, where 0.4 is the Froude Number [106]. During the acclimation trial (consisting of all eight activities), the speed was adjusted based participant verbal feedback to achieve a fixed walking speed that was a comfortable speed for all activities.

For the self-paced acclimation period, participants were given verbal instructions on how to initialize treadmill speed, increase or decrease speed, and stop. Participants practiced self-paced mode for level treadmill walking until they were confident in starting, stopping, accelerating, and decelerating. Subsequently, they completed an acclimation trial consisting of all eight walking activities.

Table 4.2. Walking activities included in the virtual scenario. Slope activities included downhill, uphill, and cross slopes. Right and left cross slope gait cycles were separated into top-cross-slope (TS) and bottom-cross-slope (BS). For right-cross-slope, the left limb was at the top of the slope and the right limb was at the bottom. For left-cross-slope, the left limb was at the bottom of the slope and the right limb was at the top. Uneven walking activities included medial-lateral platform translations, rolling-hills, and simulated rocky terrain.

Walking activity		Description
Level	Level (LW)	Level treadmill platform
Slope	Downhill (DS)	-7° decline (pitch)
	Uphill (US)	+7° incline (pitch)
	Right-cross-slope (RS)	Platform tilts to the right (roll) 5°
	Left-cross-slope (LS)	Platform tilts to the left (roll) 5°
Uneven	Medial-lateral translations (MLT)	Platform oscillates in the medial-lateral direction based on a sum of four sines with frequencies of 0.16, 0.21, 0.24, and 0.49Hz [107]. Amplitude scaling, $A_w = 0.015$, yields a maximum range of ±4 cm.
	Rolling-hills (HL)	Platform oscillates in the sagittal plane (pitch) based on a sum of four sines with frequencies of 0.16, 0.21, 0.24, and 0.49Hz [107]. Amplitude scaling, $A_w = 0.01$, yields a maximum range of ±3°.
	Rocky (RO)	Platform oscillates in three directions simultaneously using the CAREN Rumble module with a maximum range of ±2cm at 0.6 Hz vertically, ±1° at 1Hz pitch, and ±1° at 1.2Hz roll.

4.3.4 Data analyses

Marker and ground reaction force (GRF) data were filtered using a 4th order low pass Butterworth filter with 10 Hz and 20 Hz cut-off frequencies, respectively. A 13-segment body model was created in Visual3D (C-Motion Inc., Germantown, MD). The platform tracking markers defined a platform reference frame. Gait events were identified from foot velocity relative to the pelvis [108]. Treadmill speed for fixed and self-paced speed conditions was calculated for each stride using anterior-posterior 5th metatarsal marker velocity during mid-stance. During this phase, the foot and treadmill belt move at the same speed. Treadmill speed was calculated from the average of five frames with the smallest standard deviation of the velocity curve (see Appendix D for details).

Marker positions for temporal-spatial analyses were resolved in the platform reference frame. A custom Matlab program (2010a, The Mathworks, Matwick, MA) was used to calculate temporal-spatial parameters, including treadmill velocity, stride length, step length, step width, stride time, step time, terminal double support time (DST), stance time, and swing time. Double support time, stance, and swing times were represented as % of stride time. Custom Matlab programs were also developed to determine ranges-of-motion for ankle, knee, hip, pelvis, and trunk. Additionally, peak ground reaction forces were determined.

Thirty cycles were obtained for each activity, across trials. Mean and mean of the standard deviations (mSD) across 30 cycles were calculated for temporal spatial and range-of-motion parameters. For GRFs, mean and mSD across 6 cycles were calculated.

4.3.5 Statistical analyses

R version 2.14.2 was used for statistical analyses [109]. Data were examined using three statistical linear mixed models using the *lmer4* package [110]. A linear mixed model was selected since not all participants completed three trials for each speed condition. Statistical analyses were also performed with a treadmill speed covariate. Post-hoc t-tests were performed using the *multcomp* package [111] with a Holm-Bonferroni correction ($p<0.05$).

A statistical analysis was performed to determine if participants varied level walking speed throughout the trial. This analysis examined temporal-spatial parameters across eight level walking sections (excluding initial level walking) for each trial. The fixed factors were group (AB or TT), trial (1-3), limb (dominant/non-dominant or intact/prosthetic), and speed condition (fixed or self-paced). Participant was the random factor. Post-hoc comparisons were only performed to examine differences between group and speed conditions.

Since both groups significantly varied their level walking speed for the self-paced condition, further analyses were performed to determine if participants changed their temporal-spatial parameters between the preceding level section and a non-level walking activity (e.g., level section before an uphill and the uphill) and between the non-level walking activity and the following level section (e.g., uphill walking and the next level section). Only temporal-spatial parameters were examined. This second statistical analysis included group (AB or TT), activity (DS, US, TS, BS, MLT, HL, RO), limb (dominant/non-dominant or intact/prosthetic), and speed condition (fixed or self-paced) as fixed factors. Participant was the random factor. Post-hoc comparisons were only performed for speed condition.

The third statistical model examined differences between fixed and self-paced treadmill speeds during non-level walking activities. The fixed factors included group (AB or TT), activity

(DS, US, TS, BS, MLT, HL, RO), limb (dominant/non-dominant or intact/prosthetic), and speed condition (fixed or self-paced). Participant was the random factor. Temporal-spatial, range-of-motion, and peak GRF parameters were examined. Post-hoc comparisons were only performed for speed condition.

Table 4.3. Level walking mean and mean standard deviation (mSD) of treadmill speed (m/s), stride length (m), and stride time (s). Stride lengths and times are reported for able-bodied (AB) non-dominant limb and transtibial amputee (TT) prosthetic limb. Standard deviation is in brackets. Boldfaced numbers indicate significant differences between fixed and self-paced speed conditions.

			Mean		mSD	
	Parameter	Group	Fixed	Self-paced	Fixed	Self-paced
Level	Speed	AB	**1.28 (0.06)**	**1.44 (0.12)**	**0.01 (0.00)**	**0.08 (0.02)**
		TT	**1.17 (0.12)**	**1.33 (0.13)**	**0.01 (0.01)**	**0.10 (0.03)**
	Stride length	AB	**1.41 (0.09)**	**1.52 (0.11)**	**0.03 (0.01)**	**0.06 (0.01)**
		TT	**1.35 (0.13)**	**1.43 (0.13)**	**0.03 (0.01)**	**0.07 (0.02)**
	Stride time	AB	**1.10 (0.04)**	**1.06 (0.05)**	**0.016 (0.004)**	**0.024 (0.007)**
		TT	**1.17 (0.06)**	**1.11 (0.06)**	**0.025 (0.007)**	**0.038 (0.015)**

4.4 Results

4.4.1 Temporal-spatial

Treadmill speed, stride length, and stride time data presented in Table 4.3 and Table 4.4. Detailed temporal-spatial parameters can be found in Appendix E.

For the fixed-speed condition, AB and TT participants were unable to vary speed, hence they varied stride length and stride time across the eight level sections ($p<0.001$; Figure 4.1). For the self-paced speed condition, both groups significantly changed their level walking speed across the eight level sections ($p<0.001$) accompanied by changes in stride length and stride time ($p<0.001$; Table 4.3). AB and TT participants also varied their speed across the other non-level walking activities for the self-paced condition ($p<0.001$; Table 4.4). With the exception of uphill, both groups walked significantly faster for self-paced compared to the fixed-speed condition ($p<0.001$; Table 4.4). For all walking activities, speed mSD and stride length mSD were greater for self-paced compared to the fixed speed condition ($p<0.001$; Table 4.4). TT walked slower than AB for both fixed and self-paced conditions ($p<0.03$) and differences in stride length and stride time were related to treadmill speed differences between groups.

Table 4.4. Mean and mean of treadmill speed deviation (mSD) of treadmill speed (m/s), stride length (m), and stride time (s) for non-level walking activities. *During-Before* refers to the difference between the preceding level section the activity (e.g, level section before downhill and the downhill section). A positive value is an increase in the parameter (e.g., faster speed) and a negative value is a decrease in the parameter (e.g., slower speed). *During* refers to the activity (e.g., downhill). *After-During* refers to the difference between the following level section and the activity (e.g., level section after downhill and the downhill section). Stride lengths and times are reported for able-bodied (AB) non-dominant limb and transtibial amputee (TT) prosthetic limb. Standard deviation is in brackets. Boldfaced numbers indicate significant differences between fixed and self-paced speed conditions.

			During-Before		During				After-During	
			Fixed	Self-paced	Fixed	Self-paced	Fixed	Self-paced	Fixed	Self-paced
Group	Activity	Parameter	Mean	Mean	Mean	Mean	mSD	mSD	Mean	Mean
Able-bodied	Downhill	Speed	0.00 (0.00)	0.04 (0.10)	**1.29 (0.06)**	**1.45 (0.18)**	**0.01 (0.00)**	**0.09 (0.06)**	0.00 (0.01)	-0.02 (0.13)
		Stride length	-0.06 (0.03)	-0.02 (0.10)	**1.35 (0.09)**	**1.49 (0.18)**	**0.04 (0.00)**	**0.07 (0.04)**	0.06 (0.03)	0.03 (0.12)
		Stride time	-0.05 (0.02)	-0.04 (0.02)	1.05 (0.04)	1.03 (0.05)	0.02 (0.00)	0.02 (0.01)	0.05 (0.02)	0.04 (0.02)
	Uphill	Speed	**0.00 (0.00)**	**-0.16 (0.09)**	1.28 (0.06)	1.27 (0.17)	**0.01 (0.00)**	**0.10 (0.04)**	**0.00 (0.00)**	**0.15 (0.08)**
		Stride length	**0.04 (0.05)**	**-0.07 (0.08)**	1.46 (0.08)	1.44 (0.12)	**0.04 (0.01)**	**0.08 (0.03)**	**-0.03 (0.04)**	**0.07 (0.07)**
		Stride time	**0.04 (0.04)**	**0.08 (0.06)**	1.14 (0.07)	1.14 (0.09)	**0.03 (0.01)**	**0.05 (0.04)**	**-0.03 (0.04)**	**-0.06 (0.04)**
	Top-cross-slope	Speed	0.00 (0.00)	-0.01 (0.06)	**1.29 (0.06)**	**1.44 (0.15)**	**0.01 (0.00)**	**0.08 (0.03)**	0.00 (0.00)	0.02 (0.04)
		Stride length	-0.01 (0.01)	0.01 (0.02)	**1.41 (0.08)**	**1.55 (0.12)**	**0.03 (0.01)**	**0.05 (0.02)**	0.01 (0.01)	-0.01 (0.03)
		Stride time	0.00 (0.01)	0.00 (0.01)	**1.09 (0.04)**	**1.06 (0.05)**	0.02 (0.01)	0.02 (0.01)	0.00 (0.01)	0.00 (0.01)
	Bottom-cross-slope	Speed	0.00 (0.00)	0.01 (0.04)	**1.28 (0.06)**	**1.47 (0.14)**	**0.01 (0.00)**	**0.07 (0.03)**	0.00 (0.00)	-0.02 (0.07)
		Stride length	0.00 (0.02)	0.00 (0.04)	**1.41 (0.08)**	**1.53 (0.14)**	**0.03 (0.01)**	**0.06 (0.02)**	0.00 (0.01)	0.01 (0.04)
		Stride time	0.00 (0.01)	0.00 (0.01)	**1.10 (0.04)**	**1.06 (0.05)**	0.02 (0.01)	0.02 (0.01)	0.00 (0.01)	0.00 (0.01)
	Medial-lateral translations	Speed	0.00 (0.00)	0.01 (0.04)	**1.28 (0.06)**	**1.47 (0.13)**	**0.01 (0.00)**	**0.07 (0.03)**	0.00 (0.00)	-0.02 (0.05)
		Stride length	-0.03 (0.03)	-0.02 (0.03)	**1.39 (0.09)**	**1.52 (0.11)**	**0.04 (0.01)**	**0.06 (0.02)**	0.02 (0.02)	0.01 (0.04)
		Stride time	-0.02 (0.02)	-0.02 (0.02)	**1.08 (0.04)**	**1.04 (0.05)**	0.02 (0.01)	0.03 (0.01)	0.02 (0.02)	0.02 (0.02)
	Rolling hills	Speed	**0.00 (0.01)**	**-0.10 (0.11)**	1.29 (0.06)	1.32 (0.19)	**0.03 (0.01)**	**0.11 (0.04)**	**0.00 (0.01)**	**0.13 (0.08)**
		Stride length	**-0.03 (0.04)**	**-0.11 (0.09)**	1.38 (0.10)	1.40 (0.16)	**0.06 (0.02)**	**0.08 (0.02)**	**0.03 (0.04)**	**0.13 (0.07)**
		Stride time	-0.03 (0.03)	0.00 (0.03)	1.07 (0.05)	1.06 (0.08)	0.03 (0.01)	0.04 (0.02)	0.02 (0.02)	0.00 (0.03)
	Rocky	Speed	0.00 (0.01)	-0.02 (0.05)	**1.28 (0.06)**	**1.43 (0.16)**	**0.02 (0.00)**	**0.09 (0.03)**	0.00 (0.01)	0.01 (0.04)
		Stride length	-0.03 (0.05)	-0.03 (0.07)	**1.38 (0.12)**	**1.49 (0.15)**	**0.05 (0.02)**	**0.08 (0.02)**	0.04 (0.05)	0.03 (0.06)
		Stride time	-0.02 (0.03)	-0.01 (0.03)	1.08 (0.06)	1.05 (0.06)	0.03 (0.01)	0.03 (0.01)	0.02 (0.04)	0.01 (0.02)
Transtibial Amputee	Downhill	Speed	**0.00 (0.01)**	**-0.05 (0.11)**	**1.17 (0.12)**	**1.24 (0.15)**	**0.01 (0.01)**	**0.13 (0.05)**	**0.00 (0.01)**	**0.08 (0.11)**
		Stride length	**-0.12 (0.06)**	**-0.16 (0.11)**	1.24 (0.13)	1.25 (0.16)	**0.06 (0.02)**	**0.10 (0.03)**	**0.12 (0.07)**	**0.19 (0.11)**
		Stride time	-0.10 (0.05)	-0.09 (0.05)	**1.08 (0.07)**	**1.04 (0.05)**	**0.04 (0.02)**	**0.05 (0.02)**	0.09 (0.05)	0.07 (0.06)
	Uphill	Speed	**0.00 (0.00)**	**-0.13 (0.09)**	1.17 (0.12)	1.22 (0.17)	**0.01 (0.00)**	**0.10 (0.06)**	**0.00 (0.00)**	**0.10 (0.11)**
		Stride length	**0.05 (0.07)**	**-0.05 (0.06)**	1.41 (0.14)	1.40 (0.16)	**0.05 (0.02)**	**0.09 (0.05)**	**-0.03 (0.06)**	**0.03 (0.07)**
		Stride time	**0.04 (0.06)**	**0.07 (0.06)**	**1.22 (0.10)**	**1.17 (0.09)**	**0.04 (0.01)**	**0.06 (0.02)**	-0.03 (0.06)	-0.06 (0.06)
	Top-cross-slope	Speed	0.00 (0.00)	0.00 (0.04)	**1.17 (0.12)**	**1.35 (0.13)**	**0.01 (0.01)**	**0.09 (0.04)**	0.00 (0.00)	-0.01 (0.04)
		Stride length	0.00 (0.02)	0.01 (0.04)	**1.36 (0.12)**	**1.46 (0.14)**	**0.04 (0.01)**	**0.07 (0.03)**	0.00 (0.02)	0.00 (0.03)
		Stride time	0.01 (0.02)	0.00 (0.02)	**1.18 (0.06)**	**1.10 (0.06)**	0.03 (0.01)	0.04 (0.02)	-0.01 (0.01)	0.00 (0.03)
	Bottom-cross-slope	Speed	0.00 (0.00)	0.01 (0.03)	**1.17 (0.12)**	**1.36 (0.15)**	**0.01 (0.01)**	**0.09 (0.04)**	0.00 (0.01)	-0.02 (0.04)
		Stride length	0.00 (0.01)	0.00 (0.03)	**1.36 (0.13)**	**1.46 (0.14)**	**0.04 (0.01)**	**0.07 (0.03)**	0.00 (0.01)	0.00 (0.02)
		Stride time	0.00 (0.01)	-0.01 (0.02)	**1.17 (0.06)**	**1.10 (0.06)**	0.03 (0.01)	0.03 (0.01)	0.00 (0.01)	0.01 (0.02)
	Medial-lateral translations	Speed	0.00 (0.00)	0.03 (0.06)	**1.17 (0.12)**	**1.34 (0.14)**	**0.01 (0.01)**	**0.10 (0.06)**	0.00 (0.00)	0.03 (0.04)
		Stride length	-0.02 (0.02)	0.00 (0.04)	**1.33 (0.12)**	**1.43 (0.14)**	**0.05 (0.01)**	**0.08 (0.04)**	0.02 (0.02)	0.04 (0.04)
		Stride time	-0.02 (0.02)	-0.03 (0.03)	**1.16 (0.06)**	**1.08 (0.06)**	0.04 (0.01)	0.04 (0.01)	0.02 (0.02)	0.01 (0.02)
	Rolling hills	Speed	**0.00 (0.00)**	**-0.09 (0.10)**	1.18 (0.12)	1.23 (0.14)	**0.03 (0.01)**	**0.14 (0.06)**	**0.00 (0.00)**	**0.11 (0.11)**
		Stride length	**-0.02 (0.03)**	**-0.08 (0.06)**	1.33 (0.13)	1.36 (0.14)	**0.06 (0.01)**	**0.11 (0.05)**	**0.02 (0.02)**	**0.09 (0.07)**
		Stride time	**-0.02 (0.02)**	**0.02 (0.04)**	1.15 (0.06)	1.13 (0.05)	**0.04 (0.01)**	**0.08 (0.04)**	**0.02 (0.02)**	**-0.02 (0.05)**
	Rocky	Speed	0.00 (0.01)	0.00 (0.06)	**1.17 (0.12)**	**1.34 (0.11)**	**0.02 (0.00)**	**0.13 (0.06)**	0.00 (0.00)	-0.03 (0.04)
		Stride length	-0.01 (0.06)	-0.02 (0.06)	**1.34 (0.16)**	**1.43 (0.13)**	**0.07 (0.02)**	**0.11 (0.05)**	0.02 (0.06)	0.00 (0.05)
		Stride time	0.00 (0.05)	-0.01 (0.04)	**1.16 (0.08)**	**1.09 (0.06)**	0.05 (0.02)	0.05 (0.03)	0.01 (0.05)	0.02 (0.03)

For the self-paced speed condition, AB and TT participants significantly reduced walking speed (p<0.001) for uphill and rolling-hill activities, compared to the preceding level walking section, and significantly increased walking speed when returning to level (p<0.001; Table 4.4). Only TT participants significantly decreased walking speed downhill and increased walking speed when returning to level (p<0.001). Both groups maintained speed for the other non-level activities (cross-slopes, rocky, and medial-lateral translations) compared to the preceding and following level sections. When using treadmill speed as a covariate, there were no differences in temporal-spatial parameters between fixed and self-paced speed conditions for all activities except uphill and rolling-hills (p<0.002).

For uphill walking, both groups decreased stride length during self-paced trials, but increased stride length during fixed-speed trials (p<0.01), independent of treadmill speed differences between fixed and self-paced conditions. While AB and TT increased stride time similarly for uphill and for both speed conditions, the increase in stride time was greater for self-paced compared to the fixed speed condition (p<0.02).

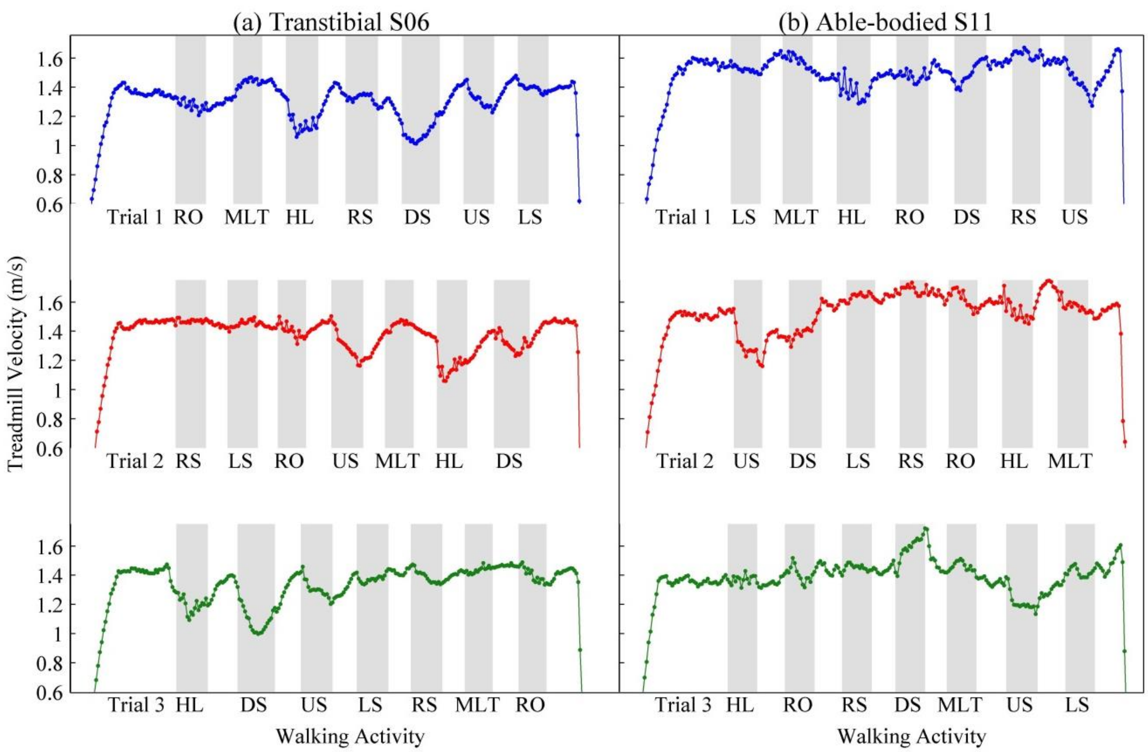

Figure 4.1. Treadmill speed for each stride for a representative transtibial amputee (a) and an able-bodied (b) participant. The shaded areas represent speed during one of the activities, and the white spaces represent speed during the level walking sections.

During the rolling-hills activity, AB and TT decreased both stride length and time during fixed speed trials, but decreased stride length and increased stride time for the self-paced condition ($p<0.001$).

4.4.2 Joint range-of-motion

Detailed range-of-motion parameters can be found in Appendix E. For both groups, hip and pelvis ranges-of-motion were significantly larger for the self-paced speed condition ($p<0.01$), compared to the fixed speed condition. However, these differences were no longer significant with a treadmill speed covariate.

When comparing range-of-motion mSD, most kinematic parameters for TT were significantly greater for self-paced compared to fixed speed ($p<0.02$), but no significant differences were found with a treadmill speed mSD covariate. AB participants also had significantly greater range-of-motion mSD for self-paced compared to the fixed speed condition, but only for hip, pelvis, and trunk parameters ($p<0.002$). When considering treadmill speed mSD as a covariate, AB pelvis and trunk mSD parameters were not significantly different between self-paced and fixed speed conditions. Additionally, when considering treadmill speed mSD as a covariate, AB ankle, knee, and hip mSD parameters were significantly smaller for self-paced compared to the fixed speed condition ($p<0.02$), which suggests that AB participants increased lower limb kinematic variability to maintain the fixed speed.

4.4.3 Ground reaction forces

Detailed peak GRF parameters can be found in Appendix E. With treadmill speed as a covariate, no mean peak GRF differences were found between speed conditions. AP-GRF peak braking mSD was significantly smaller for self-paced downhill walking, for the TT intact limb ($p<0.001$). AP-GRF peak propulsion mSD was significantly larger for self-paced on the bottom-cross-slope activity, for the TT intact limb ($p<0.001$). Although significant, these differences were less than 0.02 N/kg and may not be clinically significant.

4.5 Discussion

This research examined self-paced and fixed-paced treadmill modes when walking over various terrains within a virtual reality environment. For the self-paced treadmill mode, participants changed their walking speed due to natural variability or in response to different terrain. These speed changes were typically accompanied by changes in other temporal-spatial, kinematic, and ground reaction force parameters.

For self-paced treadmill walking, both AB and TT participants reduced treadmill speed for uphill and rolling-hill activities. TT participants also significantly reduced treadmill speed for downhill, but AB did not. Temporal-spatial parameters for the other walking activities (cross-slopes, rocky, and medial-lateral translations) were not significantly differ from the preceding level section. When walking speed was not controlled in previous studies, AB participants reduced speed uphill [26, 112, 113], downhill [112, 113], and both AB and TT participants reduced speed over uneven surfaces [33, 36, 59], compared to level walking. However, some studies reported AB participants increased speed uphill [23] and downhill [23, 26], compared to level walking. Furthermore, other studies reported AB and TT participants maintained speed uphill, downhill, and over uneven surfaces [28, 32, 73, 78, 114], compared to level walking. Differences across these studies may be affected by shorter walkways found in many facilities or may demonstrate the wide range of responses as people adapt to different terrain.

Participants did not consistently reduce speed in self-paced mode for the rocky and medial-lateral translation sections. Curtze et al. [32] examined AB and TT gait when walking over an irregular surface and reported that participants slowed down, but not significantly, possibly due to the large within group standard deviation. These results were comparable to the rocky activity results in this study. In previous research by Hak et al. [93], AB participants did not reduce speed when walking with large continuous medial-lateral perturbations on a self-paced treadmill. Medial-lateral translations in our study were smaller than Hak et al. [93] and 70% of our participants maintained speed from the preceding level section. Amplitudes for the rocky and medial-lateral translation activities may have been too small to induce a speed change in our able-bodied group and high functioning amputee group.

Generally, temporal-spatial and ranges-of-motion were related to changes in treadmill speed since differences were no longer significant with a treadmill speed covariate. However, AB and TT participants used a different stepping strategy between level and uphill for the self-paced condition compared to the fixed speed condition. For the self-paced speed condition, AB and TT participants decreased walking speed uphill by decreasing stride length and increasing stride time. In contrast, for the fixed-speed condition, both groups increased stride length and stride time to keep up with the fixed treadmill speed. Hence, when walking on a self-paced treadmill, participants naturally changed their walking speed and walking strategy when climbing a moderate slope.

Interestingly, all participants walked faster for the self-paced condition, compared to the fixed-speed condition, for all activities except uphill. Walking speed may have been underestimated for the fixed-speed condition, but all participants reported that the fixed speed was comfortable for all eight activities. Walking speed can be influenced by several factors. In the Mohler et al. [115] virtual reality-based study, if the visual scene's speed (optic flow) was slower than the treadmill speed, participant preferred walking speed increased. Faster optic flow decreased preferred walking speed. While optic flow in our study was programmed to match treadmill input speed, optic flow may have been slower than the participant's real world experience, resulting in faster self-paced treadmill speeds.

The self-paced algorithm also influenced walking speed. This algorithm adjusted treadmill speed to keep the participant at the treadmill center. However, people with conventional treadmill experience typically walk at the front, to ensure that they do not fall off the back. For self-paced mode, walking at the front of the treadmill increases speed and all participants reported having walked on a treadmill in the past. 'Untraining' of this conventional treadmill walking behavior may not have occurred, particularly as this protocol included a fixed-speed condition. Self-paced walking speed could also have been affected by other constraints such as treadmill width, length, or the safety harness (e.g., the user might increase speed if the safety harness touches their head). Future improvements to the CAREN-Extended safety bar could include a track that moves anteriorly/posteriorly to allow the participant to move more freely along the treadmill length.

Amputee walking performance also may have been affected by various prosthetic components used in this study. However, all amputee participants wore high-end energy-storing-return feet, designed for various speeds and terrain. This, combined with a repeated measures study design, minimized the effect of prosthetic components on fixed versus self-paced speed comparisons.

4.6 Conclusions

This research demonstrated AB and TT participants varied their speed when walking in a multi-terrain virtual environment on a self-paced treadmill. The largest changes in speed were observed for uphill, rolling-hills, and downhill. Both AB and TT groups responded similarly to fixed and self-paced modes for all activities except downhill, where TT participants reduced speed and AB participants did not.

The CAREN-Extended system provides a safe, controlled environment to study and to improve variable terrain walking, especially for individuals with a lower limb amputation. Differences in temporal-spatial, range-of-motion, and ground reaction force parameters between fixed and self-paced speed conditions were no longer significantly different with a speed covariate, for both AB and TT. However, when walking uphill, both groups decreased stride length during self-paced trials, and increased stride length during fixed-speed trials to maintain the constant speed. Since speed accounted for most gait variations, maintaining the ability to change walking speed in a virtual environment should result in more natural gait patterns.

Factors that affect walking speed should also be considered when interpreting results using a self-paced treadmill. Future research should investigate how optic flow affects treadmill gait at self-paced speeds for CAREN systems. Additionally, a self-paced algorithm independent of a reference position may reduce the effects on speed due to treadmill constraints or conventional treadmill walking.

4.7 Acknowledgments

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Chapter 5: Transtibial amputee and able-bodied walking strategies for maintaining stable gait in a multi-terrain virtual environment

This chapter addresses objective 3 (evaluate walking stability performance) by examining various stability metrics for AB and TT when walking in a stability-challenging VE.

The following appendices are associated with this chapter:

1. Appendix B includes a sample video of the virtual scenario.
2. Appendix D includes Matlab code for determining treadmill belt velocity from foot markers.
3. Appendix F-Appendix H include the information sheet and consent form for this research protocol.

A portion of this chapter was presented at the 4th Annual Canadian Institute for Military & Veteran Health Research Forum (CIMVHR) in Edmonton, Alberta.

Sinitski EH, Lemaire ED, Baddour N, Besemann M, Dudek NL, and Hebert JS (2013) Dynamic Stability of Able-bodied Individuals and Transtibial Amputees while Walking in a Virtual Environment . Canadian Institute for Military & Veteran Health Research Forum (CIMVHR). Edmonton, AB.

This journal manuscript was submitted for publication in 2014.

Sinitski EH, Lemaire ED, Baddour N, Besemann M, Dudek NL, Hebert JS (2014) Transtibial amputee and able-bodied walking strategies for maintaining stable gait in a multi-terrain virtual environment.

5.1 Abstract

The challenge of bipedal walking is to actively balance the body over the feet while producing forward momentum and maintaining upright posture. For amputees, maintaining stable gait poses additional challenges due to the loss of lower limb musculature and proprioception. Knowledge of gait biomechanics when walking stability is challenged is required for developing quantifiable metrics for rehabilitation assessment and intervention.

Twelve people with unilateral transtibial amputation (TT) and twelve able-bodied (AB) individuals walked in a park-like virtual environment with level, slope, and simulated uneven terrain scenarios. Walking stability performance was quantified with common stability measures, including margin-of-stability, trunk accelerations, step strategies, and gait variability.

Overall, AB and TT groups adopted similar strategies to maintain stable walking; however, stability outcome measures were greater for TT compared to AB. For more challenging conditions, AB and TT had greater trunk accelerations and walked with a more cautious and variable gait pattern by reducing velocity, increasing step width, increasing minimum foot clearance, modifying margin-of-stability, and/or increasing step variability. This study demonstrated the importance of gait variability and trunk motion measures when quantitatively assessing multi-terrain walking for people with transtibial amputations.

5.2 Introduction

Stable bipedal walking requires active balancing over the feet while maintaining forward momentum and upright posture. For individuals with a lower limb amputation, transitioning onto and off the prosthesis poses challenges due to the loss of lower limb musculature and proprioception. Therefore, amputee gait is considered more unstable than able-bodied walking and is associated with greater fall risk [11-15]. While a reasonable understanding exists about transtibial amputee gait over level ground, limited information is available on gait stability when walking over more challenging terrain.

Three approaches can be considered for analyzing walking stability. The margin-of-stability is a center-of-mass versus base-of-support (COM-BOS) method that examines body position relative to the feet [50]. Deviations outside or near the base-of-support (BOS) boundary may lead to loss of balance. A second approach examines step strategies to enhance walking stability, including a cautious step pattern with shorter, wider, slower steps, longer double support times, smaller foot-surface angles, and larger foot clearance [12, 15, 32, 34, 36, 40, 45, 47, 52, 55, 71, 73-75, 116]. A third approach examines gait variability, where greater variability suggests greater instability [12, 34, 71]. In previous research, a more cautious, variable step pattern, larger margin-of-stability, and increased trunk variability was observed for people with balance deficits when walking over a level surface, compared to a control group [12, 15, 32, 34, 36, 40, 45, 47, 52, 55, 71, 73-75, 116]. This cautious gait pattern was also observed for young adults, older adults, and lower limb amputees when walking stability was challenged [12, 15, 32, 34, 36, 40, 45, 47, 52, 55, 73, 74, 116]. Lower limb amputees also display limb asymmetries, with greater step variability on the intact limb compared to the prosthetic limb. Able-bodied gait symmetry is typically assumed, but some studies reported differences between right and left limbs that may be related to limb dominance [55, 56, 117].

The purpose of this study was to understand how able-bodied and lower limb amputees maintain stable gait when walking over a variety of stability-challenging terrains encountered in daily life. We hypothesized that amputee and able-bodied participants would walk with a more cautious, variable gait pattern for non-level conditions compared to level treadmill walking. We also hypothesized these stability measures would be greater for the amputee group compared to the able-bodied group, and both groups would display limb asymmetries with increased variability on the intact or dominant limb compared to the prosthetic or non-dominant limb.

5.3 Methods

5.3.1 Subjects

A convenience sample of twelve individuals with unilateral transtibial amputation (TT; 11 males, 1 female; mass=79.9±21.1kg; height=1.78±0.07m; age=42±10 years; Table 5.1) twelve matched able-bodied individuals (AB; 11 males, 1 female; mass=80.5±11.8kg; height=1.78±0.07m; age=39±11 years) participated in this study. AB dominant limb was defined as the preferred limb used to kick a ball. TT were a K-level of three or four [118] and wore their prosthesis daily. Participants were recruited from The Ottawa Hospital Rehabilitation Centre, Canadian Forces Health Services, Glenrose Rehabilitation Hospital, and the community. All participants provided written informed consent and the study was approved by the ethics review board at each testing institution.

5.3.2 Data collection

The CAREN-Extended virtual reality system (Motek Medical, Amsterdam, NL; Figure 3.1a, page 33) provided the controlled evaluation environment. CAREN-Extended includes a 12-camera Vicon motion capture system (Vicon Inc., Oxford, UK), 180° curved projection screen, and six degree of freedom motion platform embedded with a dual-belt instrumented treadmill (Bertec Corp., Columbus, OH). Full body kinematics were tracked using a six degree of freedom, 57 marker set [105], and platform motion was tracked with three markers. Marker data were sampled at 100 Hz. Additional anatomical landmarks were defined with a digitizing pointer (C-Motion Inc., Germantown, MD).

A 10-15 minute acclimation period consisted of self-paced treadmill training (e.g., treadmill speed automatically matches walking speed) and a warm-up trial in the virtual scenario. Pelvis tracking markers were used for self-paced treadmill speed control with a 0.7 sensitivity scaling and 1 m/s^2 maximum treadmill acceleration.

All participants completed three walking trials on a path through a virtual park scene, utilizing self-paced treadmill mode. The scene was constructed from 20m sections, representing eight walking conditions (Table 5.2). All trials started from standing, followed by a period of level walking and concluded with a level walking section. All intermediate sections were randomized and separated by level sections. The total walking distance was 340 m per trial. See Appendix B for a sample video.

Table 5.1. Transtibial amputee participant demographics.

ID	Gender	Age (yr)	Weight (kg)	Height (m)	Prosthetic side (R,L)	Etiology	Time since amputation (yr)	K-level	Prosthetic Foot
S01	M	63	63	1.83	L	Trauma	8	K4	Trias
S02	M	36	96.4	1.80	R	Trauma	5	K4	C-walk
S03	M	45	98.2	1.81	R	Trauma	8	K4	Re-Flex rotate
S04	M	35	67.3	1.73	R	Trauma	1	K4	Variflex
S05	M	44	108	1.86	L	Trauma	1	K3	Variflex
S06	M	50	109	1.85	R	Trauma	1	K4	Triton
S07	M	56	87.3	1.80	L	Trauma	22	K3	Ceterus
S08	M	45	47.3	1.73	R	Trauma	15	K4	Flex foot
S09	M	43	75.9	1.75	R	Osteosarcoma	0.67	K4	Re-Flex rotate
S10	M	37	81.8	1.80	R	Osteosarcoma	3	K4	Variflex
S11	M	29	76.4	1.78	L	Congenital	28	K4	Variflex
S12	F	27	47.7	1.60	R	Other	1	K3	Variflex

Table 5.2. Walking activities included in the park virtual scenario. Slope activities included downhill, uphill, and cross slopes. Right and left cross slope gait cycles were separated into top-cross-slope (TS) and bottom-cross-slope (BS). For right-cross-slope, the left limb was at the top of the slope and the right limb was at the bottom. For left-cross-slope, the left limb was at the bottom of the slope and the right limb was at the top. Uneven walking activities included medial-lateral platform translations, rolling-hills, and simulated rocky terrain.

Walking activity		Description
Level	Level (LW)	Level treadmill platform.
Slope	Downhill (DS)	-7° decline (pitch).
	Uphill (US)	+7° incline (pitch).
	Right-cross-slope (RS)	Platform tilts to the right (roll) 5°.
	Left-cross-slope (LS)	Platform tilts to the left (roll) 5°.
Uneven	Medial-lateral translations (MLT)	Platform oscillates in the medial-lateral direction based on a sum of four sines with frequencies of 0.16, 0.21, 0.24, and 0.49Hz [107]. Amplitude scaling, A_w = 0.015, yields a maximum range of ±4 cm.
	Rolling-hills (HL)	Platform oscillates in the sagittal plane (pitch) based on a sum of four sines with frequencies of 0.16, 0.21, 0.24, and 0.49Hz [107]. Amplitude scaling, A_w = 0.01, yields a maximum range of ±3°.
	Rocky (RO)	Platform oscillates in three directions simultaneously using the CAREN Rumble module with a maximum range of ±2cm at 0.6 Hz vertically, ±1° at 1Hz pitch, and ±1° at 1.2Hz roll.

5.3.3 Data analysis

Marker data were filtered using a 4th order low pass Butterworth filter (10 Hz). A 13-segment body model was created in Visual3D (C-Motion, Inc.) [119, 120] and platform tracking markers defined a platform reference frame. Treadmill speed was determined from anterior-posterior (AP) foot marker velocity during mid-stance by averaging five frames with the smallest standard deviation (see Appendix D for details). Thirty gait cycles were obtained for each walking activity.

Custom Matlab programs (2010a, The Mathworks, Matwick, MA) were developed to calculate the outcome measures described in Table 5.3, with mean and mean of the standard deviations (mSD) as the dependent variables. R version 2.14.2 [109] was used for all statistical analyses. Data were examined using a linear mixed model (*lmer4* package, [110]) with group (AB or

Table 5.3. Measures related to walking stability.

	Outcome Measures	Description
COM-BOS	Medial-lateral margin-of-stability (ML-MoS)	Minimum medial-lateral distance between the base-of-support boundary, defined by a lateral foot marker (5^{th} metatarsal), and the extrapolated center-of-mass during early stance [50]. Marker positions were defined relative to the lab.
Step strategies	Step width (SW)	Medial-lateral distance (m) between heel markers at foot contact. Marker positions were resolved in the platform reference frame.
	Step length (SL)	Anterior-posterior distance (m) between heel markers at foot contact. Marker positions were resolved in the platform reference frame.
	Step time (ST)	Time (s) between foot contacts.
	Double support time (DST), dominant/intact limb	Time (s) between prosthetic (non-dominant) foot contact and intact (dominant) toe off. Reported as percent of stride time.
	Double support time (DST), non-dominant/prosthetic limb	Time (s) between and intact (dominant) foot contact and prosthetic (non-dominant) toe off. Reported as percent of stride time.
	Foot angle at foot strike (FA_{FS})	Angle (deg) between the foot segment and platform at foot strike.
	Foot angle at toe off (FA_{TO})	Angle (deg) between the foot segment and platform at toe off.
	Maximum foot clearance (MaxFC)	Maximum normal distance (cm) between the foot and surface during swing.
	Minimum foot clearance (MinFC)	Minimum normal distance (cm) between the foot and surface during swing.
Variability	ML-MoS mSD	Standard deviation of ML-MoS
	Step width mSD	Standard deviation of step width
	Step length mSD	Standard deviation of step length
	Step time mSD	Standard deviation of step time
	Double support time mSD	Standard deviation of double support time
	FA_{FS} mSD	Standard deviation of foot-surface angle at foot strike
	FA_{TO} mSD	Standard deviation of foot-surface angle at toe off
	MinFC mSD	Standard deviation of MinFC
	MaxFC mSD	Standard deviation of MaxFC
	$RMS_{ML,}$ $RMS_{AP,}$ RMS_{VT}	Root-mean-square of trunk acceleration in the medial-lateral (ML), anterior-posterior (AP), and vertical (VT) directions, relative to the laboratory reference frame. Reported in units of gravity (g), by step normalized to 101 points.

TT), limb (dominant/non-dominant or intact/prosthetic), and condition as fixed factors. Participant was the random factor. Analyses were also performed with treadmill speed as a covariate. Post-hoc tests were examined using the *multcomp* package [111] with Holm-Bonferroni correction ($p<0.05$). Differences between groups, limbs, and condition versus level walking were examined. From previous studies, ML-MoS limb differences were inconsistent and have shown larger [52], smaller [32] on the prosthetic side, or no differences [57]. Therefore, a multi-regression analysis was used to determine the primary contributor to ML-MoS differences between limbs and across conditions, using maximum ML-COM velocity, ML-COM range-of-motion during stance, and step width (medial-lateral distance between foot and COM).

5.4 Results

5.4.1 Margin-of-stability

ML-MoS was not significantly different between AB and TT. ML-MoS was larger on the prosthetic or non-dominant limb, compared to the intact or dominant limb ($p<0.001$; Table 5.4). Both groups increased ML-MoS for rolling-hills and rocky conditions ($p<0.01$), but decreased ML-MoS for top-cross-slope ($p<0.01$). From the multiple regression analysis, step width and maximum ML-COM velocity significantly contributed to limb and condition differences in ML-MoS.

Table 5.4. Mean of margin-of-stability (ML-MoS). Standard deviation is in brackets. AB and TT refer to able-bodied and transtibial amputee participant groups, respectively. Significant comparisons for walking condition, compared to level (LW), with a treadmill speed covariate, are boldfaced.

			Walking Condition							
	Group	Limb	LW	DS	US	TS	BS	MLT	HL	RO
ML-MoS (cm)	AB	Dominant	7.76 (1.48)	8.03 (1.19)	8.07 (1.39)	**7.44 (1.92)**	7.86 (1.46)	8.45 (1.86)	**8.42 (1.78)**	**8.71 (1.94)**
		Non-dominant	8.16 (1.48)	8.50 (1.49)	8.31 (1.31)	**7.78 (1.75)**	8.05 (1.44)	8.50 (1.71)	**8.75 (1.64)**	**9.09 (1.75)**
	TT	Intact	7.84 (1.41)	7.91 (1.10)	8.07 (1.26)	**7.29 (1.49)**	7.92 (1.38)	7.99 (1.50)	**8.29 (1.67)**	**8.32 (1.67)**
		Prosthetic	8.17 (1.40)	8.63 (1.79)	8.77 (1.55)	**7.25 (1.38)**	8.82 (1.42)	8.30 (1.50)	**8.60 (1.72)**	**8.95 (1.81)**

5.4.1 Step strategies

TT walked significantly slower than AB for level and downhill conditions ($p<0.05$; Table 5.5), with shorter step lengths, longer double support time, and/or slower step times ($p<0.02$). Compared to level, AB significantly reduced walking speed for uphill and rolling-hills conditions ($p<0.001$) and maintained similar speeds for all other conditions. TT also slightly reduced speed for uphill, rolling-hills, and downhill, but significantly increased speed for cross-slopes, medial-lateral translations, and rocky conditions ($p<0.001$).

For uphill, AB walked slower with shorter, slower steps ($p<0.001$) whereas TT walked slower with slower steps ($p<0.03$), but did not shorten step length compared to level. AB increased step width for downhill, medial-lateral translations, rolling-hills, and rocky conditions ($p<0.003$), but TT only increased step width for uphill, rolling-hills, and rocky conditions ($p<0.02$).

AB and TT displayed limb asymmetries, with longer step times and double support times on the dominant or intact limb, compared to the non-dominant or prosthetic limb ($p<0.04$). Differences between limbs were larger for TT than AB.

Compared to the level condition, AB and TT had smaller FA_{FS} for uphill, rolling-hills, and rocky conditions ($p<0.02$), and larger FA_{FS} downhill ($p<0.001$; Table 5.5). AB and TT also had larger FA_{TO} uphill ($p<0.005$) and smaller FA_{TO} downhill ($p<0.001$). For all conditions, TT had larger FA_{FS} than AB ($p<0.005$). TT prosthetic limb also had a smaller FA_{TO} than intact limb and both AB limbs ($p<0.001$).

AB and TT maintained similar MaxFC across conditions, with the exception of downhill for TT ($p<0.001$). Compared to level, AB increased MinFC for all conditions except top-cross-slope and TT increased MinFC for all conditions except top-cross-slope and MLT ($p<0.001$). AB and TT generally had larger MaxFC on the non-dominant or prosthetic limb compared to the dominant or intact limb ($p<0.007$), but only TT had larger MinFC on the prosthetic limb compared to the intact limb ($p<0.001$).

Table 5.5. Mean of treadmill speed, step width, step length, step time, foot-surface angle at foot strike (FA_{FS}), foot-surface angle at toe off (FA_{TO}), maximum foot clearance (MaxFC), and minimum foot clearance (MinFC). Standard deviation is in brackets. AB and TT refer to able-bodied and transtibial amputee participant groups, respectively. AB limbs are dominant (D) and non-dominant (nD), and TT limbs are intact (I) and prosthetic (P). Boldfaced values highlight significant comparisons, with a treadmill speed covariate, for walking condition compared to level (LW) for all parameters except speed.

			Walking Condition							
			LW	DS	US	TS	BS	MLT	HL	RO
Speed (m/s)	AB		1.42 (0.10)	1.45 (0.18)	**1.27 (0.17)**	1.44 (0.15)	1.47 (0.14)	1.47 (0.13)	**1.32 (0.19)**	1.43 (0.16)
	TT		1.27 (0.16)	1.24 (0.15)	1.22 (0.17)	**1.35 (0.13)**	**1.36 (0.15)**	**1.34 (0.14)**	1.23 (0.14)	**1.34 (0.11)**
Step width (m)	AB	D	0.14 (0.03)	**0.15 (0.03)**	0.13 (0.04)	0.13 (0.04)	0.13 (0.04)	**0.16 (0.05)**	**0.16 (0.05)**	**0.18 (0.05)**
		nD	0.14 (0.03)	**0.15 (0.03)**	0.13 (0.03)	0.13 (0.04)	0.13 (0.04)	**0.16 (0.05)**	**0.16 (0.05)**	**0.17 (0.05)**
	TT	I	0.13 (0.04)	0.14 (0.04)	**0.14 (0.04)**	0.13 (0.04)	0.12 (0.04)	0.14 (0.04)	**0.15 (0.05)**	**0.15 (0.05)**
		P	0.13 (0.04)	0.14 (0.04)	**0.14 (0.04)**	0.12 (0.04)	0.12 (0.04)	0.14 (0.04)	**0.15 (0.05)**	**0.15 (0.05)**
Step length (m)	AB	D	0.75 (0.05)	**0.74 (0.09)**	0.72 (0.06)	0.78 (0.07)	0.76 (0.06)	0.76 (0.06)	**0.70 (0.08)**	0.75 (0.07)
		nD	0.75 (0.05)	**0.74 (0.08)**	0.71 (0.06)	0.77 (0.07)	0.76 (0.07)	0.75 (0.06)	**0.69 (0.07)**	0.74 (0.07)
	TT	I	0.71 (0.06)	**0.63 (0.10)**	**0.70 (0.09)**	0.74 (0.08)	0.74 (0.08)	0.72 (0.07)	0.69 (0.09)	0.74 (0.08)
		P	0.69 (0.07)	**0.63 (0.08)**	**0.71 (0.09)**	0.74 (0.06)	0.73 (0.08)	0.72 (0.07)	0.67 (0.09)	0.72 (0.07)
Step time (s)	AB	D	0.54 (0.02)	0.52 (0.03)	**0.57 (0.04)**	0.53 (0.03)	0.53 (0.03)	0.52 (0.03)	0.54 (0.04)	0.53 (0.03)
		nD	0.53 (0.02)	0.51 (0.02)	**0.56 (0.04)**	0.52 (0.03)	0.53 (0.03)	0.51 (0.03)	0.53 (0.04)	0.52 (0.03)
	TT	I	0.57 (0.05)	**0.53 (0.03)**	**0.58 (0.05)**	0.55 (0.03)	0.56 (0.04)	0.54 (0.04)	0.57 (0.03)	0.55 (0.04)
		P	0.56 (0.04)	**0.51 (0.03)**	**0.59 (0.04)**	0.54 (0.03)	0.54 (0.03)	0.54 (0.03)	0.56 (0.03)	0.54 (0.03)
DST (%)	AB	D	13.0 (0.7)	**11.7 (0.7)**	13.8 (0.6)	12.5 (0.5)	13.0 (0.6)	12.7 (0.6)	13.3 (0.7)	13.0 (0.6)
		nD	12.5 (0.8)	**11.2 (1.2)**	13.4 (0.7)	12.0 (0.9)	12.8 (0.9)	12.0 (0.9)	12.7 (1.2)	12.3 (0.9)
	TT	I	15.9 (2.7)	**15.3 (2.7)**	15.3 (3.3)	15.0 (2.8)	15.7 (2.9)	15.3 (2.7)	16.2 (2.8)	15.6 (2.6)
		P	12.9 (1.4)	**11.4 (2.3)**	13.0 (1.6)	12.7 (1.5)	12.8 (1.2)	12.8 (1.6)	13.1 (1.7)	12.8 (1.4)
FA_{FS} (deg)	AB	D	29.3 (3.1)	**32.7 (3.5)**	**16.9 (3.9)**	28.7 (4.4)	29.9 (3.0)	28.7 (3.5)	**23.5 (3.8)**	**27.1 (5.1)**
		nD	28.3 (2.7)	**31.8 (3.5)**	**16.8 (3.5)**	28.3 (3.1)	28.3 (3.4)	27.9 (2.5)	**22.6 (3.6)**	**26.0 (4.4)**
	TT	I	29.5 (4.0)	**31.9 (6.1)**	**18.9 (4.6)**	29.7 (5.1)	31.1 (4.1)	29.8 (4.2)	**26.3 (3.5)**	**29.5 (4.7)**
		P	29.8 (5.3)	**30.4 (5.1)**	**17.6 (8.6)**	30.8 (5.5)	30.3 (5.1)	29.7 (5.3)	**27.7 (5.5)**	**29.4 (6.0)**
FA_{TO} (deg)	AB	D	70.8 (4.9)	**63.3 (3.8)**	**72.6 (6.5)**	71.5 (5.3)	72.2 (4.2)	70.8 (4.0)	67.8 (6.0)	69.7 (5.6)
		nD	71.0 (4.6)	**63.7 (3.5)**	**72.5 (5.3)**	71.9 (3.8)	71.6 (4.2)	71.0 (3.9)	68.0 (5.3)	70.0 (4.7)
	TT	I	67.8 (5.6)	**56.5 (6.4)**	**73.3 (5.6)**	68.8 (6.1)	70.7 (4.6)	68.6 (5.9)	66.0 (6.9)	67.8 (5.9)
		P	50.7 (5.7)	47.6 (6.5)	**54.9 (5.0)**	53.8 (5.5)	51.7 (5.2)	51.8 (5.4)	50.5 (5.1)	52.3 (5.4)
MaxFC (cm)	AB	D	9.79 (1.66)	9.22 (1.55)	9.97 (1.83)	9.56 (1.59)	10.19 (1.65)	9.84 (1.60)	9.79 (1.49)	10.10 (1.53)
		nD	10.08 (1.76)	9.72 (1.79)	10.41 (2.13)	9.95 (1.73)	10.60 (1.96)	10.18 (1.91)	10.16 (1.78)	10.47 (1.80)
	TT	I	9.23 (1.49)	**7.70 (1.36)**	9.57 (1.57)	8.93 (1.44)	9.84 (1.74)	9.19 (1.57)	9.04 (1.43)	9.36 (1.54)
		P	9.29 (2.27)	**7.90 (1.68)**	10.09 (1.91)	9.28 (2.11)	10.02 (2.34)	9.65 (2.25)	9.36 (2.18)	9.79 (2.13)
MinFC (cm)	AB	D	1.29 (0.33)	**2.09 (0.84)**	**2.42 (0.79)**	1.23 (0.68)	**2.08 (0.56)**	**1.89 (0.54)**	**2.38 (0.88)**	**2.58 (0.95)**
		nD	1.30 (0.47)	**2.00 (0.72)**	**2.39 (0.80)**	1.13 (0.55)	**2.13 (0.75)**	**1.86 (0.64)**	**2.22 (0.71)**	**2.44 (0.81)**
	TT	I	1.48 (0.53)	**2.19 (0.97)**	**2.58 (0.88)**	1.04 (0.44)	**2.40 (0.75)**	1.70 (0.54)	2.02 (0.61)	**2.03 (0.60)**
		P	2.03 (0.50)	**2.86 (0.63)**	2.45 (0.64)	1.65 (0.36)	**2.86 (0.64)**	2.32 (0.54)	**2.72 (0.64)**	**2.80 (0.56)**

5.4.2 Variability

5.4.2.1 Margin-of-stability variability

TT had larger ML-MoS mSD than AB ($p<0.001$; Table 5.6). Both groups had no significant limb differences in ML-MoS mSD. TT increased ML-MoS mSD for all conditions ($p<0.04$), compared to level, but AB only increased ML-MoS mSD for medial-lateral translations, rolling-hills, and rocky conditions ($p<0.001$). From multiple regression analysis, maximum ML-COM velocity mSD, ML-COM range-of-motion mSD, and step width mSD significantly contributed to ML-MoS mSD differences between conditions, with step width mSD as the primary contributor.

Table 5.6 Mean of the standard deviations (mSD) for margin-of-stability (ML-MoS). Standard deviation is in brackets. AB and TT refer to able-bodied and transtibial amputee participant groups, respectively. Significant comparisons for walking condition, compared to level (LW), with a treadmill speed covariate, are boldfaced.

			Walking Condition							
	Group	Limb	LW	DS	US	TS	BS	MLT	HL	RO
ML-MoS (cm)	AB	Dominant	0.85 (0.19)	0.88 (0.91)	0.91 (0.21)	1.04 (0.28)	0.97 (0.19)	**2.25 (0.70)**	**1.32 (0.40)**	**1.37 (0.25)**
		Non-dominant	0.83 (0.14)	0.91 (0.12)	0.93 (0.15)	1.05 (0.32)	1.71 (2.63)	**2.34 (0.77)**	**1.19 (0.28)**	**1.25 (0.27)**
	TT	Intact	1.05 (0.17)	**1.18 (0.34)**	**1.38 (0.38)**	**1.36 (0.28)**	**1.40 (0.45)**	**2.45 (0.72)**	**1.58 (0.38)**	**1.70 (0.46)**
		Prosthetic	1.14 (0.33)	**1.52 (0.56)**	**1.50 (0.61)**	**1.49 (0.54)**	**1.39 (0.29)**	**2.48 (0.57)**	**1.43 (0.39)**	**1.74 (0.37)**

5.4.2.2 Step variability

TT was generally more variable than AB for many outcome measures and walking conditions, including mSD of step length, step width, step time, foot-surface angle, and foot clearance ($p<0.001$; Table 5.7). Compared to AB, TT had significantly greater mSD for step length for downhill, rolling-hills, and rocky ($p<0.04$), step width for all conditions except for level and MLT ($p<0.04$), and step time for downhill and rolling-hill conditions ($p<0.001$).

TT was also generally more variable on the intact limb compared to the prosthetic limb. Largest differences between TT limbs were observed for mSD of step length, step width, and foot-surface angle (Table 5.7). Only step length mSD was significantly different between AB limbs with larger step length mSD on the non-dominant limb compared to the dominant limb ($p<0.01$).

Table 5.7. Mean of standard deviations (mSD) for treadmill speed, step width, step length, step time, foot-surface angle at foot strike (FA_{FS}), foot-surface angle at toe off (FA_{TO}), maximum foot clearance (MaxFC), and minimum foot clearance (MinFC). Standard deviation is in brackets. AB and TT refer to able-bodied and transtibial amputee participant groups, respectively. AB limbs are dominant (D) and non-dominant (nD), and TT limbs are intact (I) and prosthetic (P). Boldfaced values highlight significant comparisons, with a treadmill speed covariate, for walking condition compared to level (LW) for all parameters except speed.

			Walking Condition							
			LW	DS	US	TS	BS	MLT	HL	RO
Speed (m/s)	AB		0.08 (0.03)	0.09 (0.06)	0.10 (0.04)	0.08 (0.03)	0.07 (0.03)	0.07 (0.03)	0.11 (0.04)	0.09 (0.03)
	TT		0.09 (0.04)	**0.13 (0.05)**	0.10 (0.06)	0.09 (0.04)	0.09 (0.04)	0.10 (0.06)	**0.14 (0.06)**	**0.13 (0.06)**
Step width (m)	AB	D	0.03 (0.01)	0.02 (0.00)	0.02 (0.01)	0.03 (0.01)	0.03 (0.01)	**0.07 (0.02)**	**0.03 (0.01)**	**0.04 (0.01)**
		nD	0.02 (0.01)	0.03 (0.01)	0.02 (0.01)	0.03 (0.01)	0.03 (0.01)	**0.07 (0.02)**	**0.03 (0.01)**	**0.04 (0.01)**
	TT	I	0.04 (0.01)	0.04 (0.02)	**0.05 (0.02)**	**0.05 (0.01)**	0.04 (0.01)	**0.08 (0.02)**	**0.05 (0.02)**	**0.05 (0.01)**
		P	0.03 (0.01)	0.03 (0.01)	**0.05 (0.02)**	**0.04 (0.02)**	0.04 (0.01)	**0.07 (0.02)**	**0.05 (0.02)**	**0.05 (0.02)**
Step length (m)	AB	D	0.03 (0.01)	**0.04 (0.02)**	**0.04 (0.01)**	0.03 (0.01)	0.03 (0.01)	**0.04 (0.01)**	**0.06 (0.02)**	**0.04 (0.01)**
		nD	0.04 (0.01)	**0.05 (0.02)**	**0.06 (0.02)**	0.04 (0.02)	0.04 (0.02)	**0.05 (0.02)**	**0.08 (0.03)**	**0.06 (0.02)**
	TT	I	0.05 (0.02)	**0.08 (0.02)**	**0.06 (0.02)**	0.05 (0.02)	0.06 (0.02)	**0.06 (0.04)**	**0.10 (0.04)**	**0.07 (0.03)**
		P	0.04 (0.02)	**0.06 (0.02)**	**0.06 (0.02)**	0.05 (0.03)	0.04 (0.02)	**0.05 (0.03)**	**0.08 (0.05)**	**0.07 (0.03)**
Step time (s)	AB	D	0.014 (0.003)	0.015 (0.004)	**0.026 (0.023)**	0.012 (0.003)	0.012 (0.004)	0.015 (0.006)	0.027 (0.010)	0.018 (0.003)
		nD	0.014 (0.004)	0.016 (0.005)	**0.026 (0.021)**	0.011 (0.003)	0.013 (0.004)	0.016 (0.005)	0.030 (0.007)	0.020 (0.006)
	TT	I	0.022 (0.010)	0.037 (0.014)	**0.032 (0.013)**	0.021 (0.008)	0.022 (0.009)	0.025 (0.007)	**0.047 (0.017)**	0.029 (0.014)
		P	0.018 (0.007)	0.028 (0.010)	**0.030 (0.011)**	0.020 (0.010)	0.019 (0.008)	0.020 (0.007)	**0.051 (0.038)**	0.028 (0.013)
DST (%)	AB	D	0.6 (0.2)	**0.9 (0.2)**	0.8 (0.3)	0.7 (0.1)	0.6 (0.2)	0.9 (0.2)	**1.5 (0.4)**	**0.9 (0.2)**
		nD	0.6 (0.1)	**1.0 (0.2)**	0.8 (0.2)	0.6 (0.1)	0.6 (0.2)	0.7 (0.1)	**1.5 (0.3)**	**0.9 (0.2)**
	TT	I	1.0 (0.4)	**1.7 (0.5)**	1.2 (0.3)	1.0 (0.3)	0.9 (0.3)	1.0 (0.3)	**2.0 (0.6)**	**1.3 (0.4)**
		P	0.8 (0.2)	**1.5 (0.5)**	1.1 (0.3)	0.9 (0.2)	1.0 (0.3)	1.0 (0.3)	**2.3 (0.8)**	**1.4 (0.3)**
FA_{FS} (deg)	AB	D	1.8 (0.6)	2.2 (1.0)	2.6 (0.7)	2.1 (0.8)	1.8 (0.6)	2.6 (1.1)	**5.4 (1.3)**	**3.3 (0.9)**
		nD	1.7 (0.6)	2.2 (0.9)	2.8 (0.8)	2.3 (0.7)	1.7 (0.6)	2.5 (0.5)	**4.8 (0.9)**	**3.5 (0.8)**
	TT	I	2.4 (1.0)	**4.3 (3.0)**	3.1 (1.0)	2.8 (1.0)	2.3 (0.9)	3.5 (1.6)	**5.8 (1.9)**	**4.1 (1.6)**
		P	1.9 (0.9)	2.2 (0.4)	**3.3 (1.5)**	2.0 (0.8)	1.5 (0.7)	2.3 (1.4)	**3.8 (1.2)**	3.0 (1.1)
FA_{TO} (deg)	AB	D	2.1 (0.7)	**3.5 (0.9)**	**3.3 (1.0)**	2.5 (1.3)	1.9 (0.7)	**3.7 (1.2)**	**3.6 (0.8)**	**3.8 (1.3)**
		nD	1.9 (0.6)	**3.0 (1.1)**	**2.9 (0.8)**	2.1 (0.9)	1.8 (0.5)	**3.3 (1.2)**	**3.6 (0.6)**	**3.3 (0.8)**
	TT	I	3.5 (1.3)	**4.6 (1.5)**	3.4 (1.6)	3.7 (1.1)	2.9 (0.9)	4.4 (1.6)	**4.8 (1.8)**	4.3 (1.3)
		P	1.8 (0.7)	**3.4 (0.8)**	2.3 (0.8)	2.0 (0.7)	2.0 (0.9)	2.4 (0.7)	**3.5 (1.1)**	3.2 (1.1)
MaxFC (cm)	AB	D	0.39 (0.13)	0.53 (0.15)	0.54 (0.15)	0.43 (0.12)	0.46 (0.10)	**0.63 (0.16)**	**0.65 (0.10)**	**0.69 (0.14)**
		nD	0.40 (0.10)	0.53 (0.12)	0.48 (0.13)	0.40 (0.11)	0.45 (0.13)	**0.63 (0.17)**	0.58 (0.11)	**0.60 (0.11)**
	TT	I	0.49 (0.18)	0.69 (0.14)	**0.73 (0.26)**	0.59 (0.24)	0.54 (0.17)	**0.85 (0.25)**	**0.80 (0.29)**	0.69 (0.19)
		P	0.43 (0.18)	**0.75 (0.19)**	**0.65 (0.23)**	0.49 (0.13)	0.56 (0.19)	0.50 (0.13)	**0.87 (0.42)**	**0.76 (0.30)**
MinFC (cm)	AB	D	0.29 (0.07)	**0.44 (0.11)**	**0.53 (0.28)**	**0.38 (0.20)**	**0.37 (0.10)**	**0.53 (0.12)**	**0.68 (0.19)**	**0.77 (0.21)**
		nD	0.28 (0.06)	**0.45 (0.12)**	**0.48 (0.17)**	**0.36 (0.11)**	**0.35 (0.07)**	**0.56 (0.16)**	**0.64 (0.15)**	**0.74 (0.11)**
	TT	I	0.35 (0.09)	**0.55 (0.18)**	**0.65 (0.28)**	**0.43 (0.12)**	**0.46 (0.12)**	**0.57 (0.14)**	**0.69 (0.23)**	**0.72 (0.15)**
		P	0.30 (0.09)	**0.59 (0.20)**	**0.52 (0.22)**	**0.42 (0.16)**	**0.40 (0.10)**	**0.51 (0.14)**	**0.80 (0.29)**	**0.65 (0.17)**

Both groups had greater step variability for non-level conditions. For AB and TT, step length mSD was also greater for all conditions except cross-slopes ($p<0.03$), step time mSD for uphill and rolling-hills ($p<0.02$), and step width mSD for medial-lateral translations, rolling-hills, and rocky conditions ($p<0.008$). TT also increased speed mSD for downhill, rolling-hills, and rocky ($p<0.001$), step time mSD downhill ($p<0.02$) and step width mSD for uphill and top-cross-slope ($p<0.001$).

Both groups had a larger FA_{FS} mSD for rolling-hills and rocky conditions ($p<0.01$). TT also increased FA_{FS} mSD downhill on the intact limb and uphill on the prosthetic limb ($p<0.03$). AB and TT increased FA_{TO} mSD for downhill, rolling-hills, and rocky conditions ($p<0.01$). AB also increased FA_{TO} mSD for uphill and medial-lateral translation conditions ($p<0.001$). AB and TT increased MaxFC and MinFC mSD for most conditions ($p<0.05$).

5.4.2.3 Trunk variability

Both groups had greater vertical trunk movement variability for all conditions, compared to level (RMS_{VT} $p<0.03$; Table 5.8 and Table 5.9). TT had greater AP trunk movement variability for uphill, top-cross-slope, rolling-hills, and rocky conditions (RMS_{AP} $p<0.02$), but AB only increased AP trunk variability for uphill ($p<0.001$). Both groups increased ML trunk movement variability for rocky conditions (RMS_{ML} $p<0.05$) and TT also increased RMS_{ML} uphill ($p<0.001$).

ML trunk movement variability for TT was greater than AB ($p<0.05$). TT had larger VT trunk movement when stepping onto the intact limb, compared to the prosthetic limb ($p<0.001$), but ML and AP trunk movements were generally smaller when stepping onto the intact limb ($p<0.001$). AB also exhibited limb differences with larger ML trunk movement on the dominant limb than the non-dominant limb ($p<0.03$).

Table 5.8. Mean of root-mean-square for trunk acceleration in the medial-lateral (RMS_{ML}), anterior-posterior (RMS_{AP}), and vertical (RMS_{VT}) axes for level walking (LW), medial-lateral translations (MLT), rolling-hills (HL), and rocky (RO) conditions. RMS values are in units of gravity (g). Standard deviation is in brackets. AB and TT refer to able-bodied and transtibial amputee participant groups, respectively. Only significant comparisons for walking condition, compared to level (LW), with a treadmill speed covariate, are boldfaced.

			Walking Condition			
	Group	Limb	LW	MLT	HL	RO
RHS_{ML} (g)	AB	dominant	0.087 (0.02)	0.099 (0.02)	0.093 (0.02)	**0.101 (0.02)**
		non-dominant	0.083 (0.01)	0.094 (0.02)	0.089 (0.02)	**0.098 (0.02)**
	TT	intact	0.099 (0.02)	0.109 (0.01)	0.108 (0.02)	0.113 (0.02)
		prosthetic	0.112 (0.02)	0.120 (0.02)	0.118 (0.01)	**0.125 (0.02)**
RHS_{AP} (g)	AB	dominant	0.108 (0.02)	0.110 (0.02)	0.105 (0.02)	0.112 (0.02)
		non-dominant	0.112 (0.01)	0.116 (0.01)	0.107 (0.02)	0.115 (0.02)
	TT	intact	0.110 (0.03)	0.123 (0.03)	**0.122 (0.04)**	**0.126 (0.04)**
		prosthetic	0.123 (0.04)	0.135 (0.04)	**0.126 (0.04)**	**0.136 (0.04)**
RHS_{VT} (g)	AB	dominant	0.248 (0.04)	**0.280 (0.04)**	**0.257 (0.06)**	**0.286 (0.04)**
		non-dominant	0.246 (0.03)	**0.283 (0.04)**	**0.258 (0.05)**	**0.288 (0.05)**
	TT	intact	0.247 (0.05)	**0.289 (0.06)**	**0.269 (0.07)**	**0.301 (0.05)**
		prosthetic	0.212 (0.04)	**0.249 (0.06)**	**0.235 (0.07)**	**0.260 (0.05)**

Table 5.9. Mean of root-mean-square for trunk acceleration in the medial-lateral (RMS_{ML}), anterior-posterior (RMS_{AP}), and vertical (RMS_{VT}) axes for downhill (DS), uphill (US), top-cross-slope (TS), and bottom-cross-slope (BS) conditions. RMS values are in units of gravity (g). Standard deviation is in brackets. AB and TT refer to able-bodied and transtibial amputee participant groups, respectively. Only significant comparisons for walking condition, compared to level (LW), with a treadmill speed covariate, are boldfaced.

			Walking Condition			
	Group	Limb	DS	US	TS	BS
RHS_{ML} (g)	AB	dominant	0.094 (0.02)	0.091 (0.02)	0.086 (0.02)	0.086 (0.02)
		non-dominant	0.090 (0.02)	0.088 (0.02)	0.084 (0.02)	0.080 (0.02)
	TT	intact	**0.128 (0.02)**	0.103 (0.02)	0.100 (0.01)	0.107 (0.02)
		prosthetic	0.121 (0.02)	**0.121 (0.02)**	0.117 (0.02)	0.111 (0.02)
RHS_{AP} (g)	AB	dominant	0.111 (0.02)	**0.120 (0.02)**	0.118 (0.02)	0.116 (0.02)
		non-dominant	0.111 (0.02)	**0.124 (0.01)**	0.125 (0.01)	0.117 (0.02)
	TT	intact	0.103 (0.04)	**0.132 (0.05)**	**0.131 (0.04)**	0.121 (0.03)
		prosthetic	0.109 (0.04)	**0.151 (0.05)**	**0.141 (0.04)**	0.141 (0.04)
RHS_{VT} (g)	AB	dominant	**0.344 (0.08)**	**0.274 (0.04)**	**0.267 (0.04)**	**0.283 (0.05)**
		non-dominant	**0.328 (0.08)**	**0.272 (0.04)**	**0.280 (0.04)**	**0.269 (0.04)**
	TT	intact	**0.306 (0.07)**	**0.295 (0.10)**	**0.299 (0.06)**	**0.275 (0.07)**
		prosthetic	**0.263 (0.06)**	**0.261 (0.10)**	**0.244 (0.06)**	**0.246 (0.05)**

5.5 Discussion

This research explored how able-bodied and amputee adults maintain stable gait when walking on different surface conditions, in a virtual environment. Overall, AB and TT groups adopted similar strategies to maintain stable gait; however, differences in stability outcome measures were greater for TT. For more challenging walking conditions, trunk accelerations were greater and participants adopted a more cautious and variable gait pattern by reducing velocity, increasing step width, increasing minimum foot clearance, increasing/decreasing margin-of-stability, and/or increasing step variability.

5.5.1 Margin-of-stability

ML-MoS results supported previous studies where ML-MoS changed when stability was challenged and differences were primarily related to step width [12, 52, 57, 121]. While participants increased ML-MoS for the more challenging conditions, ML-MoS was smaller for top-cross-slope. For this condition, a person may compensate for the coronal plane gravity component by reducing the moment arm about the foot and the forces needed to counteract the gravity vector. Although differences in ML-MoS were significant, the differences across all conditions were less than 1cm, as with previous studies [15, 32, 52, 55-57]. Therefore, participants made step changes to maintain ML-MoS within 1cm.

ML-MoS was not significantly different between groups. However, both AB and TT displayed limb asymmetries for ML-MoS, with larger ML-MoS on the prosthetic or non-dominant limb. From previous studies, ML-MoS limb differences were inconsistent and have shown larger [52], smaller [32] on the prosthetic side, or no differences [57]. TT often have greater lateral trunk motion on their prosthetic side; however, this greater range does not necessarily result in greater ML-COM velocity. Multiple regression analysis in this study revealed that maximum ML-COM velocity during stance significantly contributed to differences in ML-MoS between limbs. Therefore, the differences between limbs across studies for ML-MoS may be attributed to differences in maximum ML-COM velocity and/or step width. Stability parameters that include COM velocity should be considered when assessing TT gait across multiple terrains.

5.5.2 Step strategies

Both groups adapted similarly to more challenging terrain by changing speed, step width, foot-surface angle, and/or foot clearance. However, TT walked slower than AB with shorter step lengths and longer double support time. TT participants also had larger foot-surface angles at foot strike, smaller foot-surface angles at toe off, smaller maximum foot clearance, and larger minimum foot clearance. Generally, differences between groups were small and may not be clinically relevant. The largest difference between groups was smaller prosthetic foot-surface angles, which is explained by the prosthetic fixed ankle.

All participants reduced walking speed for uphill and rolling-hill conditions, accompanied by step length and step time changes. Both groups adapted to more challenging terrain similarly, except for walking uphill. Compared to level, AB participants reduced speed by 0.15 m/s, with shorter, slower steps. TT participants differed, reducing speed by 0.05 m/s with slower, wider steps, and similar step length to the level condition. Since TT participants walked significantly slower than AB on a level platform, TT may not have needed to reduce speed as much as AB to walk uphill comfortably. Interestingly, TT did not reduce step length, which would have reduced the vertical COM distance and mechanical work required for each uphill step [122, 123]. TT may have preferred to complete the uphill task as quickly as possible, with fewer steps, walking with longer, wider steps to provide the required momentum to raise the body up the slope without a powered prosthesis [124]. Alternatively, limited dorsiflexion could explain the longer step length when walking uphill, since TT participants wore energy storing feet that could not vary dorsiflexion performance. A longer step length positions the foot ahead of the knee, requiring less dorsiflexion at foot-strike.

All participants significantly increased foot clearance for conditions other than level treadmill walking, with the exception of the limb at the top-cross-slope. Compared to level, minimum foot clearance differences were less than 1.5cm, similar to walking over small obstacles [47] or with partially occluded vision [66], but much smaller than walking over real rocky surfaces [36, 59]. From group averages, AB and TT had smaller foot clearances for top-cross-slope; however, five AB and two TT participants increased foot clearance relative to level. Amputees may have less experience walking on cross slopes and may avoid this task in real life, resulting in similar swing kinematics compared to level and therefore less top foot clearance on cross-slopes (i.e., on a side slope, the ground is closer to the top foot if clearance is not increased). While all TT participants achieved unhindered swing on cross slopes, the likelihood of a stumble scenario when perturbed or on uneven cross-slope terrain is greater.

At foot strike, the foot-surface angle differed from level for downhill, uphill, rocky, and rolling-hills conditions. This was consistent with foot-surface angles reported for level and sloped surfaces in the literature [36, 39]. However, these increases or decreases were attributed to the slope rather than a stability accommodation strategy (e.g., foot angle similar to level walking, but surface angle changes).

TT participants displayed typical prosthetic gait difference between limbs, including shorter prosthetic double support time and longer step times. AB participants also exhibited these limb differences. AB and TT transitioned more quickly off the non-dominant or prosthetic limb and spent more time on the dominant or intact limb. While AB limb differences were statistically significant, these differences were smaller than TT and may not be clinically relevant.

5.5.3 Variability measures

Both AB and TT had larger trunk accelerations for non-level conditions, which were accompanied by increased step variability and ML-MoS mSD. TT gait variability was also larger than AB for many outcome measures and was greater for more walking conditions. Consistent with the literature, intact limb variability was greater than the prosthetic limb [33, 41], demonstrated for foot-surface angles, step length, and foot clearance. However, TT had larger ML and AP trunk accelerations on the prosthetic side compared to the intact side. Larger prosthetic side trunk accelerations were likely due to the challenge of trunk positioning over the prosthetic limb without normal proprioception and lower limb muscle control. AB participants also exhibited limb differences for ML and AP trunk accelerations. Although AB limb differences were smaller, ML trunk accelerations were significantly greater on the dominant side, which may have resulted in larger step length mSD observed on the non-dominant side. This demonstrated a different strategy between AB and TT groups in terms of the trunk's role in maintaining stable walking on more demanding surfaces.

5.6 Conclusions

This study examined how able-bodied and transtibial amputee populations maintain stable gait for various non-level conditions, within the CAREN-Extended virtual environment. Evaluation of center of mass versus base of support, step strategies, and gait variability showed that AB and TT groups adopted similar approaches to maintain stable gait. This included adopting a more cautious gait pattern for non-level conditions by increasing step width, ML-MoS, minimum foot clearance, and/or slower speed. Participants also increased gait variability for non-level conditions, with greater variability for TT compared to AB participants. Variability outcome measures were most sensitive to changes in walking conditions and changes between groups. Larger trunk accelerations may compromise stable gait by increasing step pattern variability to accommodate for poorer trunk control, leading to “difficult to recover” situations. Thus, minimizing trunk motion may be more important for maintaining stability for challenging environments [13, 34, 73, 125]. These results demonstrated the importance of gait variability measures in quantitative clinical assessment of stability performance, which could be used to quantify stable gait and track progress over a gait training program. Future research could develop and investigate a CAREN-based stability training protocol that focuses on minimizing trunk motion and gait variability, while enhancing biomechanical performance.

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Chapter 6: Thesis conclusions and future work

This thesis quantified capabilities of the CAREN-Extended system at The Ottawa Hospital Rehabilitation Centre and identified factors to consider when collecting data using this system. The CAREN-Extended system was also used to investigate self-paced treadmill gait and stability performance for able-bodied and transtibial amputee participants when walking in a stability-challenging environment. Each objective and its associated hypotheses are discussed below:

6.1 Objective 1: Evaluate technical aspects of the virtual environment

Hypothesis 1: Platform and treadmill operation are consistent with manufacture specifications.

Motion platform translation and rotation was consistent with manufacture specifications. Although differences between input and measured platform orientation were small, the difference between each increment was consistently slightly smaller or larger than the intended increment of 5cm or 5°. As a result, the error compounded over the total platform movement. This may affect the actual location of the force plates compared to the default location of the force plates input in the data capture software.

The results from this thesis also showed how various equipment settings affect platform acceleration. To obtain larger platform accelerations reduce the D-Flow safety filter setting or increase platform movement. Similarly, to obtain smaller platform accelerations increase the D-Flow safety filter setting or reduce platform movement. If the D-Flow safety filter setting is insufficient, use the D-Flow filter module to provide even more gradual platform motion. Understanding the acceleration characteristics of the platform is important for ensuring patient safety for people with mobility disabilities and for creating more stability-challenging environments.

Treadmill speed was not consistent with input specifications, and measured speed was 3-4% faster than the input speed. This will affect calculation of stride parameters such as step length and stride length. Additionally, these findings demonstrated optic flow or velocity of the visual scene should be scaled to match the measured speed of the treadmill.

Hypothesis 2: Platform movement affects force plate signals.

Consistent with the hypothesis, motion platform movement affected the force plate signals. Force signals increased as platform acceleration increased and were largest in the direction of platform motion. Currently, there is no accepted method to account for these effects; therefore, kinetic analyses should only process gait cycles when the platform is stationary.

Additionally, impact forces during walking also affected force plate signals on the other or unloaded force plate. Since both force plates are secured to the same base, they are not isolated from each other, and the force plates measured any vibrations through the structure (e.g., foot strike). This artifact overlaps the gait frequency components and cannot be removed completely. Therefore, larger artifacts may appear as a “transient force” in the ground reaction force signals.

Baseline drift was observed when a person walked on the treadmill. Since larger baseline drift can be expected for data collections longer than 30 minutes and force plates should be zeroed frequently when ground reaction forces are recorded or utilized in a D-Flow scenario. However, it may not be practical for the participant to get on and off the platform to reinitialize the force plate baseline. In this case, post-processing techniques can be used to account for any drift by using force data during the swing phase of gait. Appendix B provides the Visual3D pipeline used to subtract this offset.

Hypothesis 3: Treadmill operation affects force plate signals.

Consistent with the hypothesis, treadmill operation affected force plate signals. However, noise in the force signals were removed effectively using a low pass 20 Hz filter.

6.2 Objective 2: Comparison of gait biomechanics between fixed and self-paced treadmill speeds

Hypothesis 1: Gait biomechanics between fixed and self-paced treadmill speeds are not significantly different, when speed is considered.

Statistically, with speed as a covariate, temporal-spatial, joint ranges of motion, and peak ground reaction forces were not significantly different between fixed and self-paced modes. However, participants used a different stepping strategy between level and uphill for self-paced mode compared to fixed-speed mode. Participants decreased stride length, step length, stride time, step time, and walking speed uphill. For fixed-speed, participants had slightly longer, but slower steps to keep up with the fixed treadmill speed. These findings demonstrated self-paced treadmill mode is important for virtual reality systems with multiple movement scenarios in order to elicit more natural gait across walking tasks.

Hypothesis 2: Able-bodied and transtibial amputees walk slower for stability-challenging terrain compared to level treadmill walking.

Able-bodied reduced speed for uphill and rolling hills activities, and transtibial amputees reduced speed for uphill, rolling hills, and downhill activities. However, walking speed did not significantly change for cross slopes, rocky, and medial-lateral translations compared to the preceding level walking section, which was partially attributed to participants inconsistently increasing, decreasing, or maintaining speed as they did for uphill, downhill, and rolling hills.

Interestingly, participants walked faster for the self-paced condition than the fixed-speed condition. Fixed-speeds may have been underestimated, but other factors may have affected preferred walking speeds including, self-paced algorithm, optic flow, and/or constraints such as treadmill width, length, or safety harness. These factors should be considered when interpreting results using self-paced mode.

Hypothesis 3: TT walk slower for stability-challenging terrain compared to AB.

Unilateral transtibial amputees walked slower than able-bodied for all conditions. Transtibial amputee group responded similarly to self-paced treadmill mode compared to able-bodied for all conditions except downhill. Downhill is more challenging for transtibial amputees and, as a result, transtibial amputees reduced speed from the preceding level section, whereas able-bodied maintained similar speed compared to the preceding level section.

6.3 Objective 3: Assess walking stability performance when walking in a stability-challenging virtual environment

Hypothesis 1: Able-bodied and transtibial amputees exhibit significant differences in margin-of-stability, step strategies, and gait variability for non-level conditions.

Able-bodied and transtibial amputees adopted a more cautious gait pattern for non-level conditions by increasing step width, margin-of-stability, minimum foot clearance, and/or slower speed. Participants also increased gait variability for non-level conditions. Specifically, participants had larger RMS of trunk accelerations, which may have led to increased step variability.

Hypothesis 2: Transtibial amputees exhibit significant differences in margin-of-stability, step strategies, and gait variability compared to able-bodied participants.

Able-bodied and transtibial amputee groups adopted similar approaches to maintain stable gait. The largest difference between groups was observed for gait variability outcome measures, with greater step variability and RMS of trunk accelerations for transtibial amputees compared to able-bodied. Larger trunk accelerations may compromise stable gait by increasing step pattern variability and may lead to a potential fall risk situation. Minimizing upper body motion may be important for lower limb amputee gait training in environments that are more challenging than level ground.

6.4 Future work

This research contributed to the literature on immersive VE applications and the use of VE technology to understand amputee gait across multiple terrain scenarios. This research also led to new questions and areas of inquiry:

1. The CAREN-Extended system is an excellent tool for creating a variety of stability-challenging terrains. However, force plate measurements when the platform is moving do not represent ground reaction forces since the data includes forces for the person and the platform. Obtaining accurate ground reaction forces from a moving platform is needed to provide kinetic gait analysis for all virtual environments. Ideally, a real-time calibration method would be created to correct for platform acceleration effects on force plate measurements.
2. Self-paced treadmill walking allows individuals to vary walking velocity by automatically adjusting treadmill speed to match the user’s speed in real-time. While this treadmill mode could provide more realistic virtual environments, certain aspects of the current “centre of the treadmill” algorithm may result in faster or slower walking is the person drifts forward or backward on the treadmill. Self-paced training is important for users to learn how to control treadmill speed and to obtain more natural gait speed changes. Future research could investigate training methods and participant guidance to optimize the training phase. A new self-paced algorithm, independent of a reference position, should reduce the effects due to treadmill constraints or conventional treadmill walking (i.e., person used to walking at the front of a treadmill). The new algorithm could also be more natural for walking initiation and

termination. Additionally, the effect of different optic flow speeds on preferred walking speeds using the CAREN self-paced treadmill mode should be researched.

3. Variability outcome measures were most sensitive to differences between populations and walking conditions. These measures could be used to develop CAREN-based rehabilitation assessment and treatment scenarios to quantify stable gait and track progress over a gait training program. In addition, the role of the torso and upper body for maintaining stable gait should be investigated.

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Appendix A: CAREN-Extended COP test

Introduction

This CAREN-Extended test examined the difference between the location of the center of pressure (COP) and center of mass (COM). For this test, a 30 kg calibrated weight was placed at various locations on each force plate (Bertec, Corp). Ideally, the difference between the location of the COP and the location of the COM is zero. Since the force plates are embedded in motion platform, COP-COM error may be greater than the COP-COM error observed for force plates embedded in a stationary level walkway.

Methods

This test was performed several times throughout the duration of this research protocol, where test # (1) was before first data collection session, (2) 6 months after first collection and before Motek annual maintenance, (3) after Motek annual maintenance, and (4) 13 months after first data collection and after Bertec maintenance and re-calibration. All data were collected after following the standardized power-up sequence. Prior to each test, the forces plates were zeroed. Five markers were attached to the weight and were tracked using Vicon Nexus sampling at 100Hz. Force plate data were sampled at 1000Hz. A person placed the weight on the force plate and walked off the platform. Six trials of 5 s were captured for six locations on both the left (FP1) and right (FP2) force plates.

Visual3D (C-Motion, Inc) was used to calculate COP and COM of the calibrated weight. The mean medial-lateral (ML) and anterior-posterior (AP) location of COP and COM were determined for each trial.

Results

For test #1, an approximate error of 15-20 mm was observed in the ML direction on both force plates and was consistent for all locations (Table 1). The error in the AP direction was within 5 mm, but was slightly greater at the front of force plate 2 (FP2). For test #2, the error was similar to the first test. For test #3, after Motek Medical addressed the offset observed in test #1, the error in the ML direction improved, but there was still a slight offset for force plate 1 (FP1). For test #4, the error increased. Furthermore, the error was different for front, mid, and back locations on the treadmill.

Conclusion

This was a simple method used to examine differences between COP and COM of a calibrated weight. The results of this test demonstrated the importance of capturing COP data periodically to determine if a COP offset is required to improve joint moment estimates from an inverse dynamic model.

Table A1. Difference (mm) between location of the center-of-pressure COP and COM of a 30 kg calibrated weight for six locations on two force plates (FP1 and FP2). This test was performed four times over 13 months to determine an appropriate COP offset.

		Test #				Test #			
		1	2	3	4	1	2	3	4
	Location	ML	ML	ML	ML	AP	AP	AP	AP
FP1	Back Medial (R)	24.7	30.5	16.5	13.1	4.4	-5.6	-1.1	-4.8
	Back Lateral (L)	24.5	28.3	14.4	5.4	3.1	-7.7	-4.7	-3.1
	Mid Medial	22.3	26.0	14.0	10.0	3.9	-5.1	0.0	6.2
	Mid Lateral	22.3	24.5	12.7	5.3	2.3	-6.8	-2.0	6.8
	Front Medial	20.5	24.4	11.9	10.0	0.9	-2.1	3.2	18.1
	Front Lateral	21.2	22.6	11.1	3.4	2.1	-1.5	3.6	20.1
	Average	22.6	26.1	13.4	7.9	2.8	-4.8	-0.2	7.2
FP2	Back Medial (L)	15.0	18.7	5.7	4.2	-5.8	-3.0	1.4	21.1
	Back Lateral (R)	15.0	19.0	-5.5	2.4	-4.6	-5.2	0.3	22.0
	Mid Medial	15.9	16.4	6.3	4.5	-2.7	-0.9	-0.3	17.5
	Mid Lateral	13.2	16.4	1.5	2.3	-4.1	-1.1	1.4	18.9
	Front Medial	13.9	13.7	4.8	3.0	-8.1	-0.8	-4.3	12.6
	Front Lateral	13.1	15.9	5.0	3.6	-6.9	0.4	6.1	17.2
	Average	14.4	16.7	3.0	3.3	-5.4	-1.8	0.8	18.2

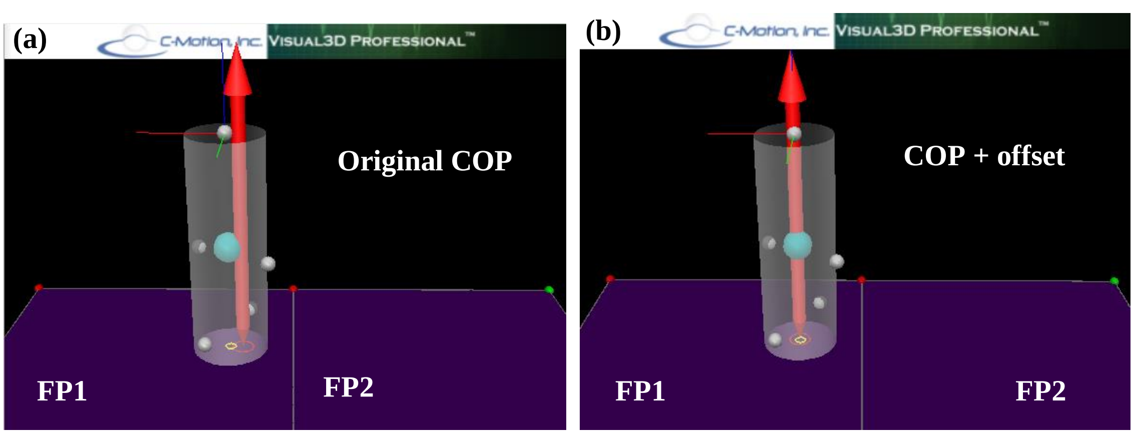

Figure A1. Original COP location (a). Adjusted COP location (b). The yellow circle represents COM of the weight projected onto the surface.

Appendix B: Video of virtual scenario

A sample video of the virtual scenario is included in the supplemental material associated with this thesis.

Appendix C: Visual3D pipeline for subtracting force offset due to platform orientation or baseline drift

The Visual3D pipeline is included in the supplemental material associated with this thesis.

This research protocol involved walking trials with both level and sloped conditions in the same motion trial. At the start of each trial, force plates were zeroed when the CAREN-Extended platform was level. This removed gravity effects from the force plate measurements in the vertical direction; however, when the platform rotated to a sloped orientation, the gravity vector acted on multiple axes of the force plate coordinate system (Figure A2). Additionally, baseline drift was observed for longer data collection sessions. Large baseline drift not only affects ground reaction forces, but will affect automatic segment-to-force assignments and automatic gait detection using force plate data. This appendix describes the method used to correct for these orientation-based forces as well as any baseline drift.

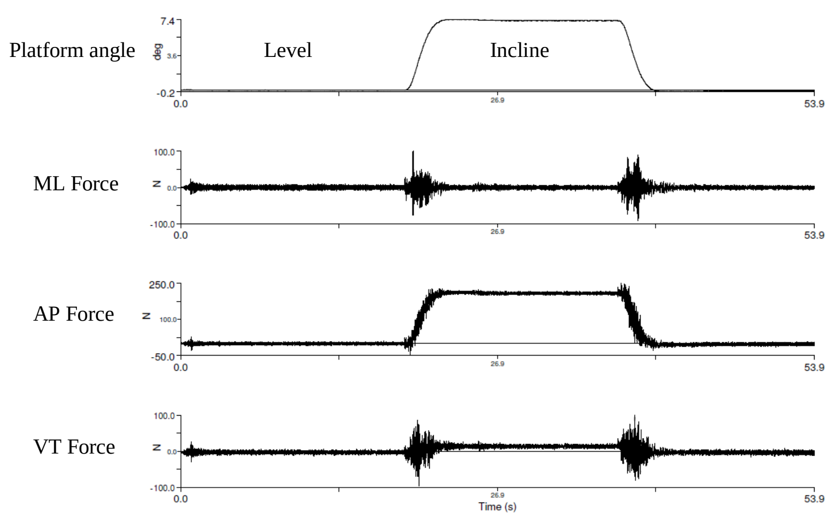

Figure A2. Platform angle, medial-lateral (ML) force, anterior-posterior (AP) force, vertical (VT) force when platform was level and 7° incline.

After a level and an incline walking c3d trial (clips) were exported from Vicon Nexus, the walking trials were loaded into Visual3D using standard laboratory pipelines. After identifying gait events, the force plate corner locations were modified for the incline walking trial using a custom pipeline and a COP offset was applied. A force offset was then determined during swing phase and subtracted from force signals. This method requires “clean” gait cycles where the left foot only contacts the left plate and the right foot only contacts the right plate. If the person steps on both force plates simultaneously with one foot, the contralateral swing phase will be affected, thus the offset signal. However, this method is acceptable because only the “clean” cycles were used for kinetic analysis.

Summary of Visual3D data processing steps

1) Run custom pipeline to load motion trials, filter target data, and create TAGs
2) Run custom pipeline to create gait events
3) Run custom pipeline to modify location of force plate corners for sloped conditions (methods not discussed here)
4) Run custom pipeline to apply COP offset (if required)
5) Run custom pipeline to determine offset and subtract offset from analog data (included as a supplemental file associated with this thesis)
    i. Create events to define a window during each swing phase (5 frames after foot off to 5 frames before foot contact);
    ii. Calculate mean during each window for each force plate signal;
    iii. Build an offset signal the same length as the original analog signal where the frames of each gait cycle correspond to the mean offset for that gait cycle;
    iv. Subtract offset signal from original analog data;
    v. Filter analog data.
6) Set to “Use Processed Analogs for Ground Reaction Force Calculations”
7) Click Recalc button
8) Identify clean or good gait cycles using a RGood or LGood event for kinetic analysis

Figure A3 shows the raw force signal, offset signal, and processed force signal calculated using the Visual3D pipeline for four gait cycles on a level platform and four gait cycles on a sloped platform at 7°.

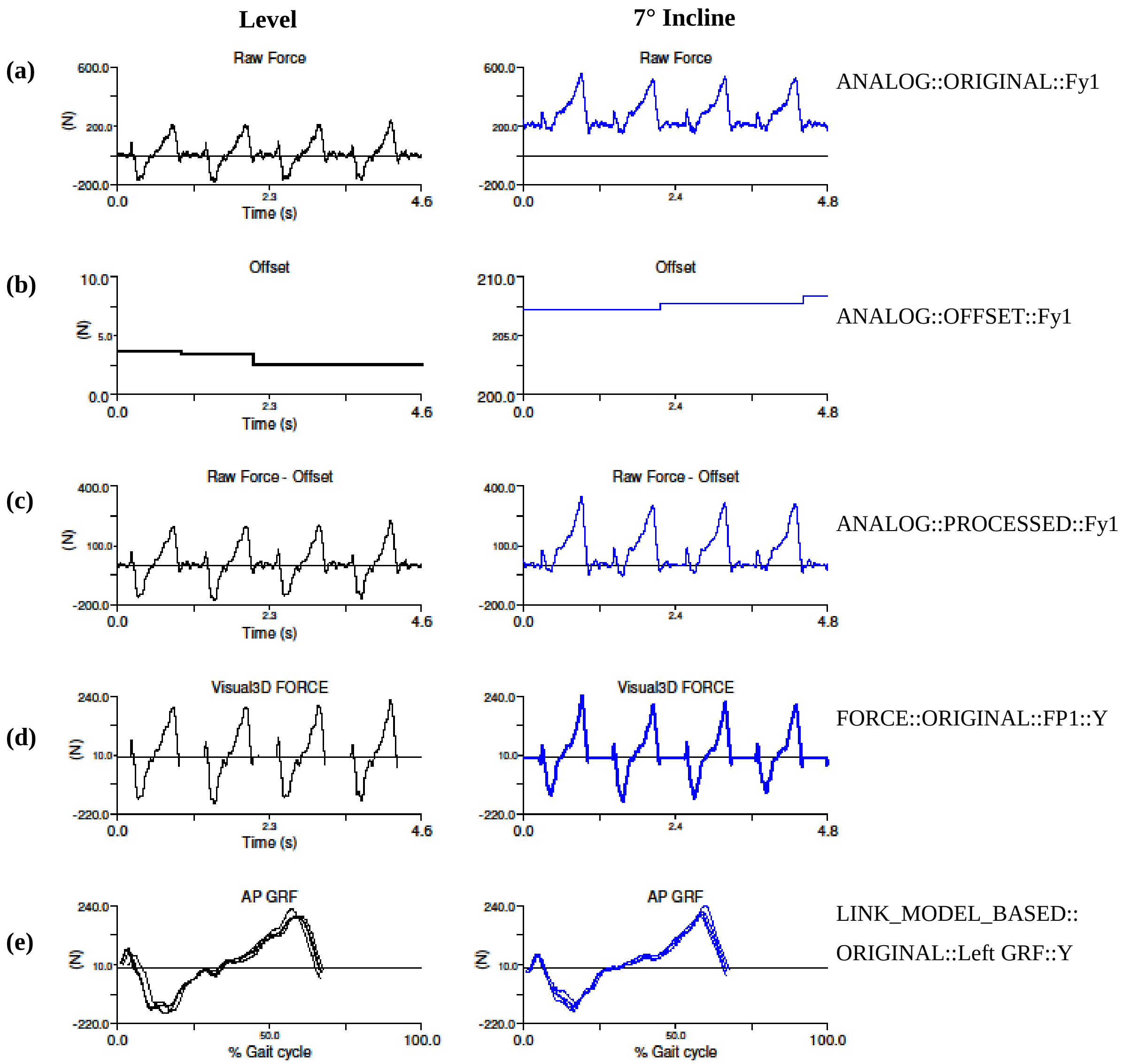

Figure A3. Original anterior-posterior force (a), offset force signal created using measured force during each swing phase (b), original force minus offset signal, filtered with low pass 20Hz (c), force signal after Visual3D recalculation to resolve force signals with respect to modified force plate corner locations (Visual3D recalculation) (d), force data plotted between foot contact gait events with a LGood intermediate event (e).

Appendix D: Matlab code for determining treadmill velocity from foot markers

This appendix describes the method used to determine treadmill velocity from foot marker anterior-posterior (AP) velocity. This approach determines a treadmill speed for each stride, which is important when using self-paced treadmill mode for calculating temporal spatial parameters, such as step length and stride length.

When walking on a dual belt treadmill (left and right treadmill belt speeds linked), the right foot is stationary relative to the right treadmill belt during the left swing phase, and the left foot is stationary relative to the left treadmill belt during the right swing phase. Therefore, AP foot marker velocity can be examined during the contralateral swing phase. The 5th metatarsal or lateral heel marker provides the longest, flattest AP velocity curve during mid-stance. Figure A4 shows AP velocity for level treadmill walking at a constant speed and Figure A5 shows AP velocity for walking on rolling hills at a self-paced speed. All marker data were resolved in the platform reference frame instead of the global or laboratory reference frame. The custom Matlab (Mathworks, Inc) function, *Calculate_TM_Velocity.m,* provided in this appendix loops through each contralateral swing phase to find the mean of five frames with the smallest standard deviation. A treadmill speed value was determined for each gait cycle is used for corresponding step length and stride length calculations.

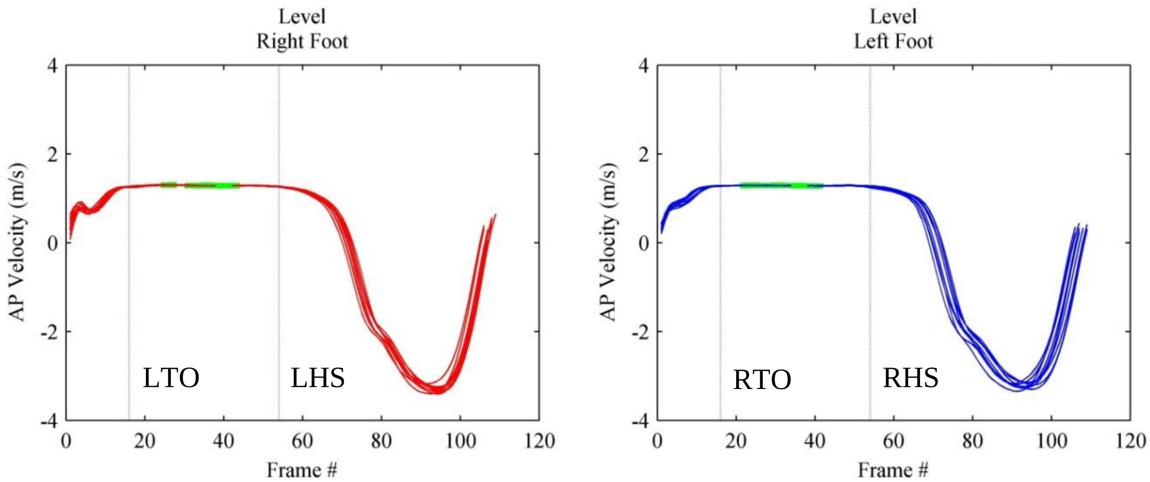

Figure A4. Right (R) and left (L) 5th metatarsal marker anterior-posterior (AP) velocity for level treadmill walking at a constant speed. The green highlighted sections represent the 5 frames with the smallest standard deviation. The vertical dotted lines represent the contralateral toe off (TO) and heel strike (HS) gait events.

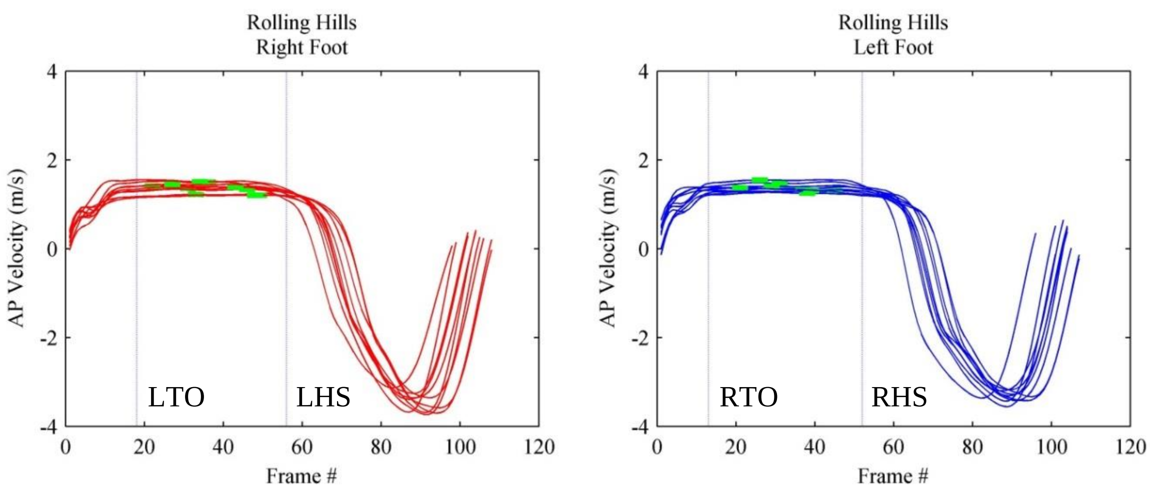

Figure A5. Right (R) and left (L) 5th metatarsal marker anterior-posterior (AP) velocity for rolling-hills condition at a self-paced speed. The green highlighted sections represent the 5 frames with the smallest standard deviation. The vertical dotted lines represent the contralateral toe off (TO) and heel strike (HS) gait events.

```matlab
%%% Sample use of the Calculate_TM_Velocity.m function.
% Rpos_DATA = right foot maker position data [X,Y,Z]
% Lpos_DATA = left foot marker position data [X,Y,Z]
% Rvel_DATA = right foot marker velocity data [X,Y,Z]
% Lvel_DATA = left foot marker velocity data [X,Y,Z]
% LHS = left heel strike gait event frame numbers
% RHS = right heel strike gait event frame numbers
% LTO = left toe off gait event frame numbers
% RTO = right toe off gait event frame numbers
% where X = medial-lateral, Y = anterior-posterior, and Z = vertical
% The number of LHS and RHS gait events should be equal
% Foot marker data should be relative to the platform, not resolved in the LAB.

[Right_TM_Vel,Left_TM_Vel] = Calculate_TM_Velocity
(Rvel_DATA(:,2),Lvel_DATA(:,2),RHS,LHS,RTO,LTO);

sampfreq = 100;
%**********************************************************
% IF the first gait event is a left heel strike
%**********************************************************
if LHS(1) < RHS(1)
   Right_Step_Width = Rpos_DATA(RHS(2:end),1) - Lpos_DATA(LHS(2:end),1);
   Right_Step_Time = (1/sampfreq)*(RHS(2:end) - LHS(2:end));
   AP = Rpos_DATA(RHS(2:end),2) - Lpos_DATA(LHS(2:end),2);
   Right_Step_Length = AP + Right_Step_Time.*Right_TM_Vel;
   Left_Step_Width = Rpos_DATA(RHS(1:end-1),1) - Lpos_DATA(LHS(2:end),1);
   Left_Step_Time = (1/sampfreq)*(LHS(2:end) - RHS(1:end-1));
   AP = Lpos_DATA(LHS(2:end),2) - Rpos_DATA(RHS(1:end-1),2);
   Left_Step_Length = AP + Left_Step_Time.*Left_TM_Vel;
%**********************************************************
% ELSE the first gait event is a right heel strike
%**********************************************************
else
   Right_Step_Width = Rpos_DATA(RHS(2:end),1) - Lpos_DATA(LHS(1:end-1),1);
   Right_Step_Time = (1/sampfreq)*(RHS(2:end) - LHS(1:end-1));
   AP = Rpos_DATA(RHS(2:end),2) - Lpos_DATA(LHS(1:end-1),2);
   Right_Step_Length = AP + Right_Step_Time.*Right_TM_Vel;
   Left_Step_Width = Rpos_DATA(RHS(2:end),1) - Lpos_DATA(LHS(2:end),1);
   Left_Step_Time = (1/sampfreq)*(LHS(2:end) - RHS(2:end));
   AP = Rpos_DATA(RHS(2:end),2) - Lpos_DATA(LHS(2:end),2);
   Left_Step_Length = AP + Left_Step_Time.*Left_TM_Vel;
end
```

```
%**********************************************************
% Calculate_TM_Velocity.m
%
% INPUTS:
% Rvel_DATA - Anterior-posterior velocity time series data for RIGHT foot
% Lvel_DATA - Anterior-posterior velocity time series data for LEFT foot
% RHS - vector of right heel strike gait event frame numbers
% LHS - vector of left heel strike gait event frame numbers
% RTO - vector of right toe off gait event frame numbers
% LTO - vector of left toe off gait event frame numbers
%
% OUTPUTS:
% Right_TM_Velocity - vector of right treadmill belt velocity, one data point
% for each step
%
% Left_TM_Velocity - vector of left treadmill belt velocity, one data point
% for each step
%
% Written by Emily Sinitski 2013
%**********************************************************
function [Right_TM_Velocity, Left_TM_Velocity] =
Calculate_TM_Velocity(Rvel_DATA,Lvel_DATA,RHS,LHS,RTO,LTO)
if length(RHS) ~= length(LHS)
 fprintf('The number of left and right cycles should be equal.\nLHS: %i\tRHS:
%i\n',length(LHS),length(RHS))
 Right_TM_Velocity = [];
 Left_TM_Velocity = [];
 return
end
 %**********************************************************
 % IF the first gait event is a left heel strike
 %**********************************************************
 if LHS(1) < RHS(1)
 %******************************************************
 % For each LTO to LHS (when the RIGHT foot is
 % stationary relative to the treadmill), calculate the standard
 % deviation (SD) of a five frame data window.
 %
 % Determine the window that has the smallest SD and
 % select the mean treadmill velocity for this window
 % to use for subsequent temporal spatial calculations.
 %******************************************************
 for i = 1:length(LTO)
 frame_num = length(Rvel_DATA(LTO(i):LHS(i+1)));
 if frame_num > 6 %make sure window is larger than 6 frames
 win_st = LTO(i); %set window start at toe off frame
 for k = 1:frame_num-4
  Right_belt_vel_sd(k) = std(Rvel_DATA(win_st:win_st+4));
  Right_belt_vel_mn(k) = mean(Rvel_DATA(win_st:win_st+4));
  win_st = win_st + 1; %set window start to next frame
 end
 [min_val,min_ind] = min(Right_belt_vel_sd);
 Right_TM_Velocity(i,1) = Right_belt_vel_mn(min_ind);
 else
 Right_TM_Velocity(i,1) = 0; %likely not a gait cycle, can be NAN
 end
 clear Right_belt_vel_sd Right_belt_vel_mn
 end
```

```
%****************************************************
% For each RTO to RHS (when the LEFT foot is
% stationary relative to the treadmill), calculate the standard
% deviation (SD) of a five frame data window.
%****************************************************
for i = 1:length(RTO)-1
frame_num = length(Lvel_DATA(RTO(i):RHS(i)));
if frame_num >6 %make sure window is larger than 6 frames
win_st = RTO(i);
for k = 1:frame_num-4
 Left_belt_vel_sd(k) = std(Lvel_DATA(win_st:win_st+4));
 Left_belt_vel_mn(k) = mean(Lvel_DATA(win_st:win_st+4));
 win_st = win_st + 1;
end
[min_val,min_ind] = min(Left_belt_vel_sd);
Left_TM_Velocity(i,1) = Left_belt_vel_mn(min_ind);
else
Left_TM_Velocity(i,1) = 0; %likely not a gait cycle, can be NAN
end
clear Left_belt_vel_sd Left_belt_vel_mn
end
%********************************************************
% ELSE the first gait event is a right heel strike
%********************************************************
else
% Determine left belt speed
for i = 1:length(RTO)
frame_num = length(Lvel_DATA(RTO(i):RHS(i+1)));
if frame_num > 6 %make sure window is larger than 6 frames
win_st = RTO(i);
for k = 1:frame_num-4
Left_belt_vel_sd(k) = std(Lvel_DATA(win_st:win_st+4));
Left_belt_vel_mn(k) = mean(Lvel_DATA(win_st:win_st+4));
win_st = win_st + 1;
end
[min_val,min_ind] = min(Left_belt_vel_sd);
Left_TM_Velocity(i,1) = Left_belt_vel_mn(min_ind);
else
Left_TM_Velocity(i,1) = 0; %likely not a gait cycle, can be NAN
end
clear Left_belt_vel_sd Left_belt_vel_mn
end
% Determine right belt speed
for i = 1:length(LTO)-1
frame_num = length(Rvel_DATA(LTO(i):LHS(i)));
if frame_num > 6 %make sure window is larger than 6 frames
win_st = LTO(i);
for k = 1:frame_num-4
Right_belt_vel_sd(k) = std(Rvel_DATA(win_st:win_st+4));
Right_belt_vel_mn(k) = mean(Rvel_DATA(win_st:win_st+4));
win_st = win_st + 1;
end
[min_val,min_ind] = min(Right_belt_vel_sd);
Right_TM_Velocity(i,1) = Right_belt_vel_mn(min_ind);
else
Right_TM_Velocity(i,1) = 0; %likely not a gait cycle, can be NAN
end
clear Right_belt_vel_sd Right_belt_vel_mn
end
end
end
```

Appendix E: Temporal-spatial, ranges-of-motion, and peak ground reaction force for fixed and self-paced speed conditions

The tables in this document present mean and mean of the standard deviations (mSD) for temporal-spatial, range-of-motion, and ground reaction force parameters for eight walking activities and two speed conditions. Standard deviation is in brackets. Boldfaced numbers indicate significant differences ($p<0.05$) between fixed and self-paced speed conditions.

Acronyms and definitions:

LW: level walking
DS: 7° decline
US: 7° incline
TS: 5° cross slope with the limb at the top of the slope
BS: 5° cross slope with the limb at the bottom of the slope
MLT: medial-lateral translations with a maximum range of ± 4 cm
HL: simulated rolling-hill terrain with a maximum pitch range of ± 3°
RO: simulated uneven (rocky) terrain with movement in three directions simultaneously

During-Before: refers to the difference between the preceding level section the activity (e.g, level section before downhill and the downhill section). A positive value is an increase in the parameter (e.g., faster speed) and a negative value is a decrease in the parameter (e.g., slower speed).

During: refers to the activity (e.g., downhill).

After-During: refers to the difference between the following level section and the activity (e.g., level section after downhill and the downhill section). A positive value is an increase in the parameter (e.g., faster speed) and a negative value is a decrease in the parameter (e.g., slower speed).

Dominant limb: limb used to kick a ball of the able-bodied group

Non-dominant limb: opposite limb to the foot used to kick a ball of the able-bodied group

Intact limb: unaffected or sound limb of the transtibial amputee group

Prosthetic limb: amputated limb with a prosthesis of the transtibial amputee group

Table A2. Temporal-spatial parameters for level walking.

		Mean		mSD	
Parameter	Group or limb	Fixed	Self-paced	Fixed	Self-paced
Speed (m/s)	Able-bodied	**1.28 (0.06)**	**1.44 (0.12)**	**0.01 (0.00)**	**0.08 (0.02)**
	Transtibial amputee	**1.17 (0.12)**	**1.33 (0.13)**	**0.01 (0.01)**	**0.10 (0.03)**
Stride length (m)	Dominant	**1.41 (0.08)**	**1.53 (0.11)**	**0.03 (0.01)**	**0.06 (0.01)**
	Non-dominant	**1.41 (0.08)**	**1.53 (0.11)**	**0.03 (0.01)**	**0.06 (0.01)**
	Intact	**1.37 (0.13)**	**1.47 (0.14)**	**0.03 (0.01)**	**0.07 (0.02)**
	Prosthetic	**1.36 (0.13)**	**1.45 (0.13)**	**0.03 (0.01)**	**0.07 (0.02)**
Step length (m)	Dominant	**0.71 (0.04)**	**0.76 (0.06)**	**0.02 (0.00)**	**0.03 (0.01)**
	Non-dominant	**0.70 (0.04)**	**0.76 (0.05)**	**0.02 (0.00)**	**0.04 (0.01)**
	Intact	**0.68 (0.07)**	**0.73 (0.07)**	**0.03 (0.01)**	**0.05 (0.01)**
	Prosthetic	**0.68 (0.06)**	**0.72 (0.07)**	**0.03 (0.01)**	**0.04 (0.02)**
Step width (m)	Dominant	0.14 (0.03)	0.14 (0.04)	**0.02 (0.01)**	**0.03 (0.01)**
	Non-dominant	0.14 (0.03)	0.14 (0.04)	**0.02 (0.01)**	**0.03 (0.01)**
	Intact	0.13 (0.04)	0.13 (0.04)	**0.04 (0.01)**	**0.04 (0.01)**
	Prosthetic	0.13 (0.04)	0.13 (0.04)	**0.03 (0.01)**	**0.04 (0.01)**
Stride time (s)	Dominant	**1.10 (0.04)**	**1.06 (0.05)**	**0.016 (0.004)**	**0.024 (0.007)**
	Non-dominant	**1.10 (0.04)**	**1.06 (0.05)**	**0.016 (0.004)**	**0.024 (0.007)**
	Intact	**1.17 (0.06)**	**1.10 (0.06)**	**0.025 (0.007)**	**0.038 (0.014)**
	Prosthetic	**1.17 (0.06)**	**1.11 (0.06)**	**0.025 (0.007)**	**0.038 (0.015)**
Step time (s)	Dominant	**0.55 (0.02)**	**0.53 (0.03)**	0.010 (0.002)	0.014 (0.004)
	Non-dominant	**0.55 (0.02)**	**0.53 (0.02)**	0.010 (0.002)	0.013 (0.003)
	Intact	**0.59 (0.03)**	**0.56 (0.04)**	0.017 (0.005)	0.023 (0.007)
	Prosthetic	**0.58 (0.03)**	**0.55 (0.03)**	0.014 (0.004)	0.019 (0.007)
DST (%)	Dominant	**13.44 (0.55)**	**12.90 (0.59)**	0.58 (0.07)	0.65 (0.10)
	Non-dominant	**12.88 (0.77)**	**12.33 (0.96)**	0.60 (0.10)	0.66 (0.10)
	Intact	**16.38 (2.45)**	**15.45 (2.72)**	**0.92 (0.18)**	**1.06 (0.23)**
	Prosthetic	**13.47 (1.26)**	**12.76 (1.35)**	**0.80 (0.21)**	**0.92 (0.11)**
Stance (%)	Dominant	**63.12 (0.67)**	**62.57 (0.75)**	0.60 (0.11)	0.65 (0.09)
	Non-dominant	**63.21 (0.55)**	**62.67 (0.73)**	0.61 (0.10)	0.68 (0.10)
	Intact	**65.78 (2.31)**	**64.95 (2.65)**	0.93 (0.21)	0.99 (0.17)
	Prosthetic	**64.06 (1.52)**	**63.27 (1.72)**	0.86 (0.20)	0.96 (0.18)
Swing (%)	Dominant	**36.88 (0.67)**	**37.43 (0.75)**	0.60 (0.11)	0.65 (0.09)
	Non-dominant	**36.79 (0.55)**	**37.33 (0.73)**	0.61 (0.10)	0.68 (0.10)
	Intact	**34.22 (2.31)**	**35.05 (2.65)**	0.93 (0.21)	0.99 (0.17)
	Prosthetic	**35.94 (1.52)**	**36.73 (1.72)**	0.86 (0.20)	0.96 (0.18)

Table A3. Mean and mean of standard deviations (mSD) for treadmill speed.

Treadmill speed (m/s)

		During-Before		During				After-During	
		Fixed	Self-paced	Fixed	Self-paced	Fixed	Self-paced	Fixed	Self-paced
Activity	Limb	Mean	Mean	Mean	Mean	mSD	mSD	Mean	Mean
DS	Able-bodied	0.00 (0.00)	0.04 (0.10)	**1.29 (0.06)**	**1.45 (0.18)**	**0.01 (0.00)**	**0.09 (0.06)**	0.00 (0.00)	-0.02 (0.13)
	Transtibial amputee	0.00 (0.00)	-0.05 (0.11)	**1.17 (0.12)**	**1.24 (0.15)**	**0.01 (0.01)**	**0.13 (0.05)**	0.00 (0.00)	0.08 (0.11)
US	Able-bodied	**0.00 (0.00)**	**-0.16 (0.09)**	1.28 (0.06)	1.27 (0.17)	**0.01 (0.00)**	**0.10 (0.04)**	**0.00 (0.00)**	**0.15 (0.08)**
	Transtibial amputee	**0.00 (0.00)**	**-0.13 (0.09)**	1.17 (0.12)	1.22 (0.17)	**0.01 (0.00)**	**0.10 (0.06)**	**0.00 (0.00)**	**0.10 (0.11)**
TS	Able-bodied	0.00 (0.00)	-0.01 (0.06)	**1.29 (0.06)**	**1.44 (0.15)**	**0.01 (0.00)**	**0.08 (0.03)**	0.00 (0.00)	0.02 (0.04)
	Transtibial amputee	0.00 (0.00)	0.00 (0.04)	**1.17 (0.12)**	**1.35 (0.13)**	**0.01 (0.01)**	**0.09 (0.04)**	0.00 (0.00)	-0.01 (0.04)
BS	Able-bodied	0.00 (0.00)	0.01 (0.04)	**1.28 (0.06)**	**1.47 (0.14)**	**0.01 (0.00)**	**0.07 (0.03)**	0.00 (0.00)	-0.02 (0.07)
	Transtibial amputee	0.00 (0.00)	0.01 (0.03)	**1.17 (0.12)**	**1.36 (0.15)**	**0.01 (0.01)**	**0.09 (0.04)**	0.00 (0.00)	-0.02 (0.04)
MLT	Able-bodied	0.00 (0.00)	0.01 (0.04)	**1.28 (0.06)**	**1.47 (0.13)**	**0.01 (0.00)**	**0.07 (0.03)**	0.00 (0.00)	-0.02 (0.05)
	Transtibial amputee	0.00 (0.00)	0.03 (0.06)	**1.17 (0.12)**	**1.34 (0.14)**	**0.01 (0.01)**	**0.10 (0.06)**	0.00 (0.00)	0.03 (0.04)
HL	Able-bodied	**0.00 (0.00)**	**-0.10 (0.11)**	1.29 (0.06)	1.32 (0.19)	**0.03 (0.01)**	**0.11 (0.04)**	**0.00 (0.00)**	**0.13 (0.08)**
	Transtibial amputee	**0.00 (0.00)**	**-0.09 (0.10)**	1.18 (0.12)	1.23 (0.14)	**0.03 (0.01)**	**0.14 (0.06)**	**0.00 (0.00)**	**0.11 (0.11)**
RO	Able-bodied	0.00 (0.00)	-0.02 (0.05)	**1.28 (0.06)**	**1.43 (0.16)**	**0.02 (0.00)**	**0.09 (0.03)**	0.00 (0.00)	0.01 (0.04)
	Transtibial amputee	0.00 (0.00)	0.00 (0.06)	**1.17 (0.12)**	**1.34 (0.11)**	**0.02 (0.00)**	**0.13 (0.06)**	0.00 (0.00)	-0.03 (0.04)

Table A4. Mean and mean of standard deviations (mSD) for stride length.

Stride Length (m)

		During-Before		During				After-During	
		Fixed	Self-paced	Fixed	Self-paced	Fixed	Self-paced	Fixed	Self-paced
Activity	Limb	Mean	Mean	Mean	Mean	mSD	mSD	Mean	Mean
DS	Dominant	-0.05 (0.03)	-0.02 (0.10)	**1.35 (0.09)**	**1.49 (0.18)**	**0.03 (0.01)**	**0.08 (0.04)**	0.06 (0.03)	0.03 (0.12)
	Non-dominant	-0.06 (0.03)	-0.02 (0.10)	**1.35 (0.09)**	**1.49 (0.18)**	**0.04 (0.00)**	**0.07 (0.04)**	0.06 (0.03)	0.03 (0.12)
	Intact	**-0.10 (0.06)**	**-0.15 (0.11)**	1.27 (0.14)	1.28 (0.17)	**0.05 (0.02)**	**0.10 (0.03)**	**0.11 (0.07)**	**0.18 (0.11)**
	Prosthetic	**-0.12 (0.06)**	**-0.16 (0.11)**	1.24 (0.13)	1.25 (0.16)	**0.06 (0.02)**	**0.10 (0.03)**	**0.12 (0.07)**	**0.19 (0.11)**
US	Dominant	**0.04 (0.05)**	**-0.08 (0.08)**	1.46 (0.08)	1.44 (0.12)	**0.04 (0.01)**	**0.08 (0.03)**	**-0.03 (0.04)**	**0.07 (0.07)**
	Non-dominant	**0.04 (0.05)**	**-0.07 (0.08)**	1.46 (0.08)	1.44 (0.12)	**0.04 (0.01)**	**0.08 (0.03)**	**-0.03 (0.04)**	**0.07 (0.07)**
	Intact	**0.05 (0.07)**	**-0.05 (0.06)**	1.43 (0.14)	1.42 (0.17)	**0.05 (0.01)**	**0.09 (0.05)**	**-0.03 (0.06)**	**0.03 (0.07)**
	Prosthetic	**0.05 (0.07)**	**-0.05 (0.06)**	1.41 (0.14)	1.40 (0.16)	**0.05 (0.02)**	**0.09 (0.05)**	**-0.03 (0.06)**	**0.03 (0.07)**
TS	Dominant	0.00 (0.02)	0.00 (0.05)	**1.41 (0.08)**	**1.53 (0.14)**	**0.03 (0.01)**	**0.06 (0.02)**	0.00 (0.02)	0.01 (0.04)
	Non-dominant	-0.01 (0.01)	0.01 (0.02)	**1.41 (0.08)**	**1.55 (0.12)**	**0.03 (0.01)**	**0.05 (0.02)**	0.01 (0.01)	-0.01 (0.03)
	Intact	0.00 (0.01)	0.00 (0.03)	**1.37 (0.13)**	**1.48 (0.15)**	**0.04 (0.01)**	**0.07 (0.03)**	0.01 (0.01)	0.00 (0.04)
	Prosthetic	0.00 (0.02)	0.01 (0.04)	**1.36 (0.12)**	**1.46 (0.14)**	**0.04 (0.01)**	**0.07 (0.03)**	0.00 (0.02)	0.00 (0.03)
BS	Dominant	-0.01 (0.01)	0.01 (0.03)	**1.40 (0.08)**	**1.55 (0.11)**	**0.03 (0.01)**	**0.05 (0.02)**	0.01 (0.01)	-0.02 (0.05)
	Non-dominant	0.00 (0.02)	0.00 (0.04)	**1.41 (0.08)**	**1.53 (0.14)**	**0.03 (0.01)**	**0.06 (0.02)**	0.00 (0.01)	0.01 (0.04)
	Intact	0.00 (0.02)	0.02 (0.04)	**1.37 (0.12)**	**1.48 (0.14)**	**0.04 (0.01)**	**0.07 (0.03)**	0.00 (0.03)	-0.01 (0.05)
	Prosthetic	0.00 (0.01)	0.00 (0.03)	**1.36 (0.13)**	**1.46 (0.14)**	**0.04 (0.01)**	**0.07 (0.03)**	0.00 (0.01)	0.00 (0.02)
MLT	Dominant	-0.03 (0.02)	-0.02 (0.03)	**1.39 (0.09)**	**1.52 (0.11)**	**0.04 (0.01)**	**0.06 (0.02)**	0.02 (0.02)	0.01 (0.04)
	Non-dominant	-0.03 (0.03)	-0.02 (0.03)	**1.39 (0.09)**	**1.52 (0.11)**	**0.04 (0.01)**	**0.06 (0.02)**	0.02 (0.02)	0.01 (0.04)
	Intact	-0.03 (0.02)	0.00 (0.04)	**1.34 (0.12)**	**1.45 (0.15)**	**0.05 (0.01)**	**0.08 (0.04)**	0.03 (0.02)	0.05 (0.04)
	Prosthetic	-0.02 (0.02)	0.00 (0.04)	**1.33 (0.12)**	**1.43 (0.14)**	**0.05 (0.01)**	**0.08 (0.04)**	0.02 (0.02)	0.04 (0.04)
HL	Dominant	**-0.03 (0.04)**	**-0.11 (0.09)**	1.38 (0.10)	1.40 (0.16)	**0.06 (0.01)**	**0.08 (0.02)**	**0.02 (0.04)**	**0.13 (0.06)**
	Non-dominant	**-0.03 (0.04)**	**-0.11 (0.09)**	1.38 (0.10)	1.40 (0.16)	**0.06 (0.02)**	**0.08 (0.02)**	**0.03 (0.04)**	**0.13 (0.07)**
	Intact	**-0.02 (0.03)**	**-0.07 (0.06)**	1.35 (0.13)	1.38 (0.15)	**0.07 (0.01)**	**0.11 (0.04)**	**0.02 (0.02)**	**0.09 (0.07)**
	Prosthetic	**-0.02 (0.03)**	**-0.08 (0.06)**	1.33 (0.13)	1.36 (0.14)	**0.06 (0.01)**	**0.11 (0.05)**	**0.02 (0.02)**	**0.09 (0.07)**
RO	Dominant	-0.03 (0.05)	-0.03 (0.07)	**1.38 (0.12)**	**1.49 (0.15)**	**0.05 (0.01)**	**0.07 (0.02)**	0.04 (0.05)	0.03 (0.06)
	Non-dominant	-0.03 (0.05)	-0.03 (0.07)	**1.38 (0.12)**	**1.49 (0.15)**	**0.05 (0.02)**	**0.08 (0.02)**	0.04 (0.05)	0.03 (0.06)
	Intact	-0.01 (0.06)	-0.02 (0.06)	**1.35 (0.16)**	**1.45 (0.14)**	**0.07 (0.02)**	**0.10 (0.05)**	0.02 (0.05)	-0.01 (0.04)
	Prosthetic	-0.01 (0.06)	-0.02 (0.06)	**1.34 (0.16)**	**1.43 (0.13)**	**0.07 (0.02)**	**0.11 (0.05)**	0.02 (0.06)	0.00 (0.05)

Table A5. Mean and mean of standard deviations (mSD) for step length.

Step Length (m)

		During-Before		During				After-During	
		Fixed	Self-paced	Fixed	Self-paced	Fixed	Self-paced	Fixed	Self-paced
Activity	Limb	Mean	Mean	Mean	Mean	mSD	mSD	Mean	Mean
DS	Dominant	-0.03 (0.02)	-0.01 (0.05)	**0.68 (0.05)**	**0.74 (0.09)**	**0.02 (0.01)**	**0.04 (0.02)**	0.03 (0.02)	0.01 (0.06)
	Non-dominant	-0.03 (0.01)	-0.01 (0.06)	**0.68 (0.04)**	**0.74 (0.08)**	**0.03 (0.00)**	**0.05 (0.02)**	0.03 (0.02)	0.01 (0.07)
	Intact	-0.05 (0.07)	-0.08 (0.07)	0.63 (0.08)	0.63 (0.10)	**0.05 (0.02)**	**0.08 (0.02)**	**0.05 (0.06)**	**0.10 (0.07)**
	Prosthetic	-0.07 (0.04)	-0.09 (0.06)	0.61 (0.07)	0.63 (0.08)	**0.04 (0.02)**	**0.06 (0.02)**	**0.07 (0.04)**	**0.09 (0.05)**
US	Dominant	**0.02 (0.03)**	**-0.04 (0.05)**	0.73 (0.03)	0.72 (0.06)	**0.03 (0.01)**	**0.04 (0.01)**	**-0.01 (0.03)**	**0.04 (0.04)**
	Non-dominant	**0.02 (0.04)**	**-0.04 (0.04)**	0.72 (0.05)	0.71 (0.06)	**0.03 (0.01)**	**0.06 (0.02)**	**-0.02 (0.03)**	**0.05 (0.05)**
	Intact	**0.02 (0.05)**	**-0.03 (0.03)**	0.71 (0.08)	0.70 (0.09)	**0.05 (0.02)**	**0.06 (0.02)**	**-0.02 (0.05)**	**0.03 (0.04)**
	Prosthetic	**0.03 (0.05)**	**-0.01 (0.05)**	0.72 (0.07)	0.71 (0.09)	**0.04 (0.01)**	**0.06 (0.02)**	**-0.03 (0.05)**	**0.00 (0.06)**
TS	Dominant	0.02 (0.02)	0.01 (0.02)	**0.72 (0.03)**	**0.78 (0.07)**	0.02 (0.01)	0.03 (0.01)	-0.01 (0.02)	-0.01 (0.02)
	Non-dominant	-0.01 (0.01)	0.00 (0.04)	**0.69 (0.04)**	**0.77 (0.07)**	0.03 (0.01)	0.04 (0.02)	0.01 (0.02)	-0.01 (0.03)
	Intact	0.01 (0.02)	0.00 (0.02)	**0.69 (0.08)**	**0.74 (0.08)**	0.03 (0.01)	0.05 (0.02)	0.00 (0.03)	0.00 (0.03)
	Prosthetic	0.01 (0.03)	0.02 (0.04)	**0.69 (0.06)**	**0.74 (0.06)**	0.04 (0.02)	0.05 (0.03)	-0.01 (0.03)	0.00 (0.03)
BS	Dominant	-0.02 (0.01)	-0.01 (0.02)	**0.69 (0.04)**	**0.76 (0.06)**	0.01 (0.01)	0.03 (0.01)	0.02 (0.01)	0.01 (0.03)
	Non-dominant	0.01 (0.03)	0.00 (0.04)	**0.71 (0.03)**	**0.76 (0.07)**	0.03 (0.01)	0.04 (0.02)	-0.01 (0.02)	0.00 (0.03)
	Intact	0.01 (0.05)	0.01 (0.02)	**0.69 (0.07)**	**0.74 (0.08)**	**0.04 (0.01)**	**0.06 (0.02)**	-0.01 (0.04)	-0.01 (0.04)
	Prosthetic	-0.01 (0.03)	0.00 (0.03)	**0.67 (0.07)**	**0.73 (0.08)**	**0.03 (0.01)**	**0.04 (0.02)**	0.01 (0.03)	0.01 (0.02)
MLT	Dominant	-0.01 (0.01)	-0.01 (0.02)	**0.70 (0.04)**	**0.76 (0.06)**	0.03 (0.01)	0.04 (0.01)	0.01 (0.01)	0.00 (0.03)
	Non-dominant	-0.02 (0.02)	-0.01 (0.02)	**0.69 (0.05)**	**0.75 (0.06)**	0.03 (0.01)	0.05 (0.02)	0.01 (0.02)	0.00 (0.02)
	Intact	-0.02 (0.02)	0.00 (0.02)	**0.67 (0.07)**	**0.72 (0.07)**	**0.04 (0.01)**	**0.06 (0.04)**	0.01 (0.02)	0.02 (0.03)
	Prosthetic	0.00 (0.02)	0.00 (0.03)	**0.67 (0.07)**	**0.72 (0.07)**	**0.03 (0.01)**	**0.05 (0.03)**	0.01 (0.03)	0.02 (0.02)
HL	Dominant	**-0.01 (0.02)**	**-0.05 (0.05)**	0.69 (0.05)	0.70 (0.08)	**0.04 (0.01)**	**0.06 (0.02)**	**0.01 (0.02)**	**0.06 (0.04)**
	Non-dominant	**-0.02 (0.02)**	**-0.06 (0.04)**	0.68 (0.05)	0.69 (0.07)	**0.06 (0.02)**	**0.08 (0.03)**	**0.02 (0.02)**	**0.07 (0.04)**
	Intact	**-0.01 (0.02)**	**-0.03 (0.04)**	0.67 (0.07)	0.69 (0.09)	**0.06 (0.02)**	**0.10 (0.04)**	**0.01 (0.02)**	**0.05 (0.04)**
	Prosthetic	**-0.01 (0.02)**	**-0.04 (0.04)**	0.66 (0.07)	0.67 (0.09)	**0.05 (0.01)**	**0.08 (0.05)**	**0.01 (0.02)**	**0.05 (0.05)**
RO	Dominant	-0.02 (0.03)	-0.01 (0.04)	**0.69 (0.05)**	**0.75 (0.07)**	0.04 (0.01)	0.04 (0.01)	0.02 (0.03)	0.01 (0.03)
	Non-dominant	-0.01 (0.03)	-0.02 (0.04)	**0.69 (0.05)**	**0.74 (0.07)**	0.04 (0.01)	0.06 (0.02)	0.02 (0.03)	0.02 (0.04)
	Intact	0.00 (0.03)	0.00 (0.04)	**0.68 (0.08)**	**0.74 (0.08)**	**0.06 (0.01)**	**0.07 (0.03)**	0.01 (0.04)	-0.02 (0.04)
	Prosthetic	0.00 (0.04)	-0.01 (0.03)	**0.68 (0.08)**	**0.72 (0.07)**	**0.05 (0.01)**	**0.07 (0.03)**	0.00 (0.05)	-0.01 (0.02)

Table A6. Mean and mean of standard deviations (mSD) for step width.

Step Width (m)

		During-Before		During				After-During	
		Fixed	Self-paced	Fixed	Self-paced	Fixed	Self-paced	Fixed	Self-paced
Activity	Limb	Mean	Mean	Mean	Mean	mSD	mSD	Mean	Mean
DS	Dominant	0.02 (0.02)	0.02 (0.02)	0.16 (0.03)	0.15 (0.03)	0.02 (0.01)	0.02 (0.00)	-0.02 (0.01)	-0.02 (0.01)
	Non-dominant	0.02 (0.02)	0.02 (0.01)	0.16 (0.03)	0.15 (0.03)	0.02 (0.01)	0.03 (0.01)	-0.02 (0.01)	-0.02 (0.01)
	Intact	0.01 (0.02)	0.00 (0.01)	0.14 (0.05)	0.14 (0.04)	0.04 (0.02)	0.04 (0.02)	-0.01 (0.02)	-0.01 (0.01)
	Prosthetic	0.00 (0.02)	0.01 (0.01)	0.14 (0.05)	0.14 (0.04)	0.03 (0.02)	0.03 (0.01)	0.00 (0.02)	-0.01 (0.02)
US	Dominant	-0.01 (0.02)	0.00 (0.01)	0.13 (0.04)	0.13 (0.04)	0.03 (0.01)	0.02 (0.01)	0.01 (0.01)	0.00 (0.02)
	Non-dominant	-0.01 (0.01)	0.00 (0.01)	0.13 (0.04)	0.13 (0.03)	0.03 (0.01)	0.02 (0.01)	0.01 (0.01)	0.00 (0.02)
	Intact	0.01 (0.01)	0.01 (0.02)	0.15 (0.05)	0.14 (0.04)	0.05 (0.02)	0.05 (0.02)	-0.01 (0.02)	-0.01 (0.02)
	Prosthetic	0.01 (0.02)	0.01 (0.02)	0.15 (0.05)	0.14 (0.04)	0.05 (0.02)	0.05 (0.02)	-0.01 (0.02)	-0.01 (0.02)
TS	Dominant	-0.01 (0.01)	0.00 (0.02)	0.13 (0.04)	0.13 (0.04)	0.03 (0.01)	0.03 (0.01)	0.01 (0.01)	0.00 (0.02)
	Non-dominant	0.00 (0.02)	-0.01 (0.02)	0.14 (0.04)	0.13 (0.04)	0.03 (0.01)	0.03 (0.01)	0.01 (0.01)	0.00 (0.02)
	Intact	-0.01 (0.01)	0.00 (0.02)	0.13 (0.04)	0.13 (0.04)	0.04 (0.01)	0.05 (0.01)	0.00 (0.02)	0.00 (0.02)
	Prosthetic	-0.01 (0.02)	-0.01 (0.01)	0.13 (0.05)	0.12 (0.04)	0.04 (0.01)	0.04 (0.02)	0.01 (0.02)	0.00 (0.02)
BS	Dominant	0.00 (0.02)	-0.01 (0.02)	0.14 (0.04)	0.13 (0.04)	0.03 (0.01)	0.03 (0.01)	0.00 (0.01)	0.00 (0.02)
	Non-dominant	-0.01 (0.01)	-0.01 (0.02)	0.13 (0.04)	0.13 (0.04)	0.03 (0.01)	0.03 (0.01)	0.00 (0.01)	0.00 (0.02)
	Intact	-0.01 (0.02)	-0.01 (0.01)	0.13 (0.05)	0.12 (0.04)	0.05 (0.01)	0.04 (0.01)	0.00 (0.01)	0.01 (0.01)
	Prosthetic	-0.01 (0.01)	0.00 (0.02)	0.13 (0.04)	0.12 (0.04)	0.04 (0.01)	0.04 (0.01)	0.00 (0.01)	0.00 (0.01)
MLT	Dominant	0.02 (0.01)	0.02 (0.02)	0.16 (0.05)	0.16 (0.05)	0.06 (0.02)	0.07 (0.02)	-0.02 (0.02)	-0.02 (0.02)
	Non-dominant	0.02 (0.01)	0.02 (0.02)	0.17 (0.04)	0.16 (0.05)	0.06 (0.02)	0.07 (0.02)	-0.02 (0.02)	-0.02 (0.02)
	Intact	0.02 (0.01)	0.01 (0.01)	0.15 (0.04)	0.14 (0.04)	0.08 (0.02)	0.08 (0.02)	-0.01 (0.01)	-0.01 (0.01)
	Prosthetic	0.01 (0.01)	0.01 (0.01)	0.15 (0.05)	0.14 (0.04)	0.08 (0.02)	0.07 (0.02)	-0.01 (0.01)	-0.01 (0.01)
HL	Dominant	0.01 (0.01)	0.02 (0.01)	0.16 (0.04)	0.16 (0.05)	0.03 (0.01)	0.03 (0.01)	-0.01 (0.01)	-0.02 (0.01)
	Non-dominant	0.01 (0.01)	0.02 (0.02)	0.16 (0.04)	0.16 (0.05)	0.03 (0.01)	0.03 (0.01)	-0.01 (0.01)	-0.02 (0.01)
	Intact	0.01 (0.01)	0.02 (0.01)	0.15 (0.05)	0.15 (0.05)	0.05 (0.01)	0.05 (0.02)	-0.01 (0.01)	-0.02 (0.02)
	Prosthetic	0.01 (0.01)	0.02 (0.01)	0.15 (0.05)	0.15 (0.05)	0.05 (0.02)	0.05 (0.02)	-0.01 (0.01)	-0.02 (0.02)
RO	Dominant	0.03 (0.02)	0.03 (0.03)	0.17 (0.05)	0.18 (0.05)	0.04 (0.02)	0.04 (0.01)	-0.02 (0.02)	-0.03 (0.03)
	Non-dominant	0.03 (0.02)	0.03 (0.02)	0.17 (0.05)	0.17 (0.05)	0.04 (0.01)	0.04 (0.01)	-0.02 (0.02)	-0.03 (0.03)
	Intact	0.02 (0.01)	0.02 (0.02)	0.16 (0.05)	0.15 (0.05)	0.06 (0.02)	0.05 (0.01)	-0.02 (0.01)	-0.02 (0.02)
	Prosthetic	0.02 (0.01)	0.02 (0.02)	0.16 (0.05)	0.15 (0.05)	0.05 (0.02)	0.05 (0.02)	-0.02 (0.01)	-0.02 (0.02)

Table A7. Mean and mean of standard deviations (mSD) for stride time.

Stride Time (s)

		During-Before		During				After-During	
		Fixed	Self-paced	Fixed	Self-paced	Fixed	Self-paced	Fixed	Self-paced
Activity	Limb	Mean	Mean	Mean	Mean	mSD	mSD	Mean	Mean
DS	Dominant	-0.05 (0.02)	-0.04 (0.02)	1.05 (0.05)	1.03 (0.05)	0.02 (0.00)	0.02 (0.01)	0.05 (0.02)	0.04 (0.02)
	Non-dominant	-0.05 (0.02)	-0.04 (0.02)	1.05 (0.04)	1.03 (0.05)	0.02 (0.00)	0.02 (0.01)	0.05 (0.02)	0.04 (0.02)
	Intact	-0.10 (0.05)	-0.09 (0.05)	**1.08 (0.07)**	**1.04 (0.05)**	**0.04 (0.02)**	**0.06 (0.02)**	0.09 (0.05)	0.07 (0.06)
	Prosthetic	-0.10 (0.05)	-0.09 (0.05)	**1.08 (0.07)**	**1.04 (0.05)**	**0.04 (0.02)**	**0.05 (0.02)**	0.09 (0.05)	0.07 (0.06)
US	Dominant	**0.04 (0.04)**	**0.08 (0.06)**	1.14 (0.07)	1.14 (0.09)	**0.03 (0.01)**	**0.05 (0.05)**	**-0.04 (0.04)**	**-0.06 (0.05)**
	Non-dominant	**0.04 (0.04)**	**0.08 (0.06)**	1.14 (0.07)	1.14 (0.09)	**0.03 (0.01)**	**0.05 (0.04)**	**-0.03 (0.04)**	**-0.06 (0.04)**
	Intact	**0.04 (0.06)**	**0.07 (0.06)**	**1.22 (0.10)**	**1.17 (0.09)**	**0.04 (0.01)**	**0.06 (0.02)**	-0.03 (0.06)	-0.06 (0.06)
	Prosthetic	**0.04 (0.06)**	**0.07 (0.06)**	**1.22 (0.10)**	**1.17 (0.09)**	**0.04 (0.01)**	**0.06 (0.02)**	-0.03 (0.06)	-0.06 (0.06)
TS	Dominant	0.00 (0.01)	0.00 (0.01)	**1.10 (0.04)**	**1.06 (0.05)**	0.02 (0.01)	0.02 (0.01)	0.00 (0.01)	-0.01 (0.01)
	Non-dominant	0.00 (0.01)	0.00 (0.01)	**1.09 (0.04)**	**1.06 (0.05)**	0.02 (0.01)	0.02 (0.01)	0.00 (0.01)	0.00 (0.01)
	Intact	0.00 (0.01)	-0.01 (0.02)	**1.17 (0.06)**	**1.10 (0.06)**	0.03 (0.01)	0.03 (0.01)	0.00 (0.01)	0.00 (0.02)
	Prosthetic	0.01 (0.02)	0.00 (0.02)	**1.18 (0.06)**	**1.10 (0.06)**	0.03 (0.01)	0.04 (0.02)	-0.01 (0.01)	0.00 (0.03)
BS	Dominant	0.00 (0.01)	0.00 (0.01)	**1.09 (0.04)**	**1.06 (0.05)**	0.02 (0.01)	0.02 (0.01)	0.00 (0.01)	0.00 (0.01)
	Non-dominant	0.00 (0.01)	0.00 (0.01)	**1.10 (0.04)**	**1.06 (0.05)**	0.02 (0.01)	0.02 (0.01)	0.00 (0.01)	0.00 (0.01)
	Intact	0.01 (0.02)	0.00 (0.02)	**1.18 (0.06)**	**1.10 (0.06)**	0.03 (0.01)	0.04 (0.02)	-0.01 (0.02)	0.01 (0.02)
	Prosthetic	0.00 (0.01)	-0.01 (0.02)	**1.17 (0.06)**	**1.10 (0.06)**	0.03 (0.01)	0.03 (0.01)	0.00 (0.01)	0.01 (0.02)
MLT	Dominant	-0.02 (0.02)	-0.02 (0.02)	**1.08 (0.04)**	**1.04 (0.05)**	0.02 (0.01)	0.03 (0.01)	0.02 (0.02)	0.02 (0.02)
	Non-dominant	-0.02 (0.02)	-0.02 (0.02)	**1.08 (0.04)**	**1.04 (0.05)**	0.02 (0.01)	0.03 (0.01)	0.02 (0.02)	0.02 (0.02)
	Intact	-0.02 (0.02)	-0.03 (0.03)	**1.15 (0.06)**	**1.08 (0.06)**	0.03 (0.01)	0.04 (0.01)	0.02 (0.02)	0.01 (0.02)
	Prosthetic	-0.02 (0.02)	-0.03 (0.03)	**1.16 (0.06)**	**1.08 (0.06)**	0.04 (0.01)	0.04 (0.01)	0.02 (0.02)	0.01 (0.02)
HL	Dominant	-0.03 (0.03)	0.00 (0.03)	1.07 (0.05)	1.06 (0.08)	0.03 (0.01)	0.04 (0.01)	0.02 (0.02)	0.00 (0.03)
	Non-dominant	-0.03 (0.03)	0.00 (0.03)	1.07 (0.05)	1.06 (0.08)	0.03 (0.01)	0.04 (0.02)	0.02 (0.02)	0.00 (0.03)
	Intact	**-0.02 (0.02)**	**0.02 (0.04)**	1.15 (0.06)	1.13 (0.05)	**0.04 (0.01)**	**0.08 (0.05)**	**0.02 (0.02)**	**-0.02 (0.05)**
	Prosthetic	**-0.02 (0.02)**	**0.02 (0.04)**	1.15 (0.06)	1.13 (0.05)	**0.04 (0.01)**	**0.08 (0.04)**	**0.02 (0.02)**	**-0.02 (0.05)**
RO	Dominant	-0.02 (0.03)	-0.01 (0.02)	1.08 (0.06)	1.05 (0.06)	0.03 (0.01)	0.03 (0.01)	0.02 (0.03)	0.01 (0.02)
	Non-dominant	-0.02 (0.03)	-0.01 (0.03)	1.08 (0.06)	1.05 (0.06)	0.03 (0.01)	0.03 (0.01)	0.02 (0.04)	0.01 (0.02)
	Intact	0.00 (0.05)	-0.01 (0.04)	**1.16 (0.08)**	**1.09 (0.06)**	0.05 (0.02)	0.05 (0.03)	0.01 (0.05)	0.02 (0.03)
	Prosthetic	0.00 (0.05)	-0.01 (0.04)	**1.16 (0.08)**	**1.09 (0.06)**	0.05 (0.02)	0.05 (0.03)	0.01 (0.05)	0.02 (0.03)

Table A8. Mean and mean of standard deviations (mSD) for step time.

Step Time (s)

		During-Before		During				After-During	
		Fixed	Self-paced	Fixed	Self-paced	Fixed	Self-paced	Fixed	Self-paced
Activity	Limb	Mean	Mean	Mean	Mean	mSD	mSD	Mean	Mean
DS	Dominant	-0.03 (0.02)	-0.02 (0.01)	0.53 (0.03)	0.52 (0.03)	0.01 (0.00)	0.01 (0.00)	0.03 (0.02)	0.02 (0.01)
	Non-dominant	-0.02 (0.01)	-0.02 (0.01)	0.52 (0.02)	0.51 (0.02)	0.01 (0.00)	0.02 (0.00)	0.02 (0.01)	0.02 (0.01)
	Intact	-0.05 (0.04)	-0.04 (0.03)	**0.55 (0.04)**	**0.53 (0.03)**	0.03 (0.02)	0.04 (0.01)	0.04 (0.04)	0.04 (0.03)
	Prosthetic	-0.05 (0.02)	-0.05 (0.03)	**0.53 (0.03)**	**0.51 (0.03)**	0.02 (0.01)	0.03 (0.01)	0.05 (0.02)	0.04 (0.03)
US	Dominant	**0.02 (0.02)**	**0.04 (0.03)**	0.57 (0.03)	0.57 (0.04)	**0.02 (0.00)**	**0.03 (0.02)**	**-0.02 (0.02)**	**-0.03 (0.02)**
	Non-dominant	**0.02 (0.02)**	**0.04 (0.03)**	0.57 (0.04)	0.56 (0.04)	**0.02 (0.00)**	**0.03 (0.02)**	**-0.02 (0.02)**	**-0.03 (0.02)**
	Intact	**0.01 (0.03)**	**0.03 (0.03)**	**0.61 (0.05)**	**0.58 (0.05)**	**0.02 (0.01)**	**0.03 (0.01)**	-0.01 (0.03)	-0.02 (0.03)
	Prosthetic	**0.03 (0.03)**	**0.04 (0.03)**	**0.61 (0.05)**	**0.59 (0.04)**	**0.02 (0.01)**	**0.03 (0.01)**	-0.03 (0.03)	-0.04 (0.03)
TS	Dominant	0.00 (0.01)	0.00 (0.01)	**0.55 (0.02)**	**0.53 (0.03)**	0.01 (0.00)	0.01 (0.00)	0.00 (0.01)	0.00 (0.01)
	Non-dominant	0.00 (0.01)	0.00 (0.00)	**0.54 (0.02)**	**0.52 (0.03)**	0.01 (0.00)	0.01 (0.00)	0.00 (0.01)	0.00 (0.01)
	Intact	0.00 (0.01)	-0.01 (0.01)	**0.59 (0.03)**	**0.55 (0.03)**	0.02 (0.01)	0.02 (0.01)	0.01 (0.01)	0.00 (0.01)
	Prosthetic	0.00 (0.01)	0.00 (0.01)	**0.58 (0.04)**	**0.54 (0.03)**	0.02 (0.01)	0.02 (0.01)	0.00 (0.01)	0.00 (0.01)
BS	Dominant	0.00 (0.01)	0.00 (0.01)	**0.55 (0.02)**	**0.53 (0.03)**	0.01 (0.00)	0.01 (0.00)	0.00 (0.01)	0.00 (0.01)
	Non-dominant	0.01 (0.01)	0.01 (0.01)	**0.55 (0.02)**	**0.53 (0.03)**	0.01 (0.00)	0.01 (0.00)	0.00 (0.01)	-0.01 (0.01)
	Intact	0.01 (0.01)	0.00 (0.01)	**0.60 (0.03)**	**0.56 (0.04)**	0.02 (0.01)	0.02 (0.01)	-0.01 (0.01)	0.00 (0.01)
	Prosthetic	0.01 (0.01)	0.00 (0.01)	**0.58 (0.03)**	**0.54 (0.03)**	0.02 (0.01)	0.02 (0.01)	0.00 (0.01)	0.00 (0.01)
MLT	Dominant	-0.01 (0.01)	-0.01 (0.01)	**0.54 (0.02)**	**0.52 (0.03)**	0.01 (0.00)	0.02 (0.01)	0.01 (0.01)	0.01 (0.01)
	Non-dominant	-0.01 (0.01)	-0.01 (0.01)	**0.54 (0.02)**	**0.51 (0.03)**	0.02 (0.01)	0.02 (0.00)	0.01 (0.01)	0.01 (0.01)
	Intact	-0.02 (0.01)	-0.02 (0.01)	**0.58 (0.03)**	**0.54 (0.04)**	0.03 (0.01)	0.02 (0.01)	0.01 (0.01)	0.01 (0.01)
	Prosthetic	0.00 (0.01)	-0.01 (0.01)	**0.58 (0.03)**	**0.54 (0.03)**	0.02 (0.00)	0.02 (0.01)	0.00 (0.01)	0.00 (0.01)
HL	Dominant	-0.01 (0.01)	0.00 (0.02)	0.54 (0.03)	0.54 (0.04)	**0.02 (0.01)**	**0.03 (0.01)**	0.01 (0.01)	0.00 (0.02)
	Non-dominant	-0.01 (0.01)	0.00 (0.02)	0.53 (0.03)	0.53 (0.04)	**0.02 (0.01)**	**0.03 (0.01)**	0.01 (0.01)	0.00 (0.02)
	Intact	**-0.01 (0.01)**	**0.01 (0.02)**	0.58 (0.03)	0.57 (0.03)	**0.03 (0.01)**	**0.05 (0.02)**	**0.01 (0.01)**	**-0.01 (0.02)**
	Prosthetic	**-0.01 (0.01)**	**0.01 (0.03)**	0.57 (0.04)	0.56 (0.03)	**0.03 (0.01)**	**0.05 (0.04)**	**0.01 (0.01)**	**-0.01 (0.03)**
RO	Dominant	-0.01 (0.02)	0.00 (0.01)	0.54 (0.03)	0.53 (0.03)	0.02 (0.00)	0.02 (0.00)	0.01 (0.02)	0.01 (0.01)
	Non-dominant	-0.01 (0.02)	0.00 (0.01)	0.53 (0.03)	0.52 (0.03)	0.02 (0.01)	0.02 (0.01)	0.01 (0.02)	0.01 (0.01)
	Intact	0.00 (0.03)	0.00 (0.02)	**0.59 (0.04)**	**0.55 (0.04)**	0.03 (0.01)	0.03 (0.01)	0.01 (0.03)	0.01 (0.02)
	Prosthetic	0.00 (0.02)	0.00 (0.02)	**0.58 (0.04)**	**0.54 (0.03)**	0.03 (0.01)	0.03 (0.01)	0.00 (0.02)	0.01 (0.02)

Table A9. Mean and mean of standard deviations (mSD) for terminal double support time.

Double support time (% stride time)

		During-Before		During				After-During	
		Fixed	Self-paced	Fixed	Self-paced	Fixed	Self-paced	Fixed	Self-paced
Activity	Limb	Mean	Mean	Mean	Mean	mSD	mSD	Mean	Mean
DS	Dominant	-1.17 (0.59)	-1.33 (0.41)	12.3 (0.7)	11.7 (0.7)	0.8 (0.1)	0.9 (0.2)	1.14 (0.58)	1.30 (0.50)
	Non-dominant	-1.07 (0.74)	-1.19 (0.74)	11.7 (1.2)	11.2 (1.2)	0.8 (0.2)	1.0 (0.2)	1.15 (0.76)	1.24 (0.77)
	Intact	-0.64 (1.40)	-0.44 (1.17)	15.6 (2.3)	15.3 (2.7)	1.7 (0.6)	1.7 (0.5)	0.76 (1.43)	0.32 (0.98)
	Prosthetic	-1.53 (1.95)	-1.53 (1.85)	11.7 (2.0)	11.4 (2.3)	1.6 (0.7)	1.5 (0.5)	1.45 (2.04)	1.32 (1.88)
US	Dominant	**0.42 (0.54)**	**0.89 (0.63)**	13.8 (0.5)	13.8 (0.6)	0.7 (0.1)	0.8 (0.3)	-0.37 (0.50)	-0.88 (0.63)
	Non-dominant	**0.36 (0.84)**	**1.02 (0.66)**	13.3 (0.6)	13.4 (0.7)	0.7 (0.2)	0.8 (0.2)	-0.57 (0.78)	-1.02 (0.71)
	Intact	**-0.72 (1.24)**	**-0.05 (1.45)**	15.7 (2.7)	15.3 (3.3)	1.1 (0.4)	1.2 (0.3)	0.63 (1.29)	0.18 (1.60)
	Prosthetic	**0.09 (0.76)**	**0.28 (1.19)**	13.6 (1.5)	13.0 (1.6)	1.1 (0.4)	1.1 (0.3)	-0.13 (1.02)	-0.27 (1.22)
TS	Dominant	-0.30 (0.29)	-0.34 (0.37)	13.1 (0.7)	12.5 (0.5)	0.6 (0.1)	0.7 (0.1)	0.29 (0.29)	0.32 (0.35)
	Non-dominant	-0.34 (0.37)	-0.26 (0.57)	12.5 (0.8)	12.0 (0.9)	0.6 (0.1)	0.6 (0.1)	0.38 (0.38)	0.24 (0.61)
	Intact	-0.41 (0.52)	-0.18 (0.70)	15.9 (2.4)	15.0 (2.8)	1.0 (0.3)	1.0 (0.3)	0.40 (0.45)	0.29 (0.56)
	Prosthetic	-0.31 (0.67)	-0.10 (0.34)	13.1 (1.6)	12.7 (1.5)	0.9 (0.4)	0.9 (0.2)	0.41 (0.67)	0.06 (0.31)
BS	Dominant	0.10 (0.28)	0.11 (0.28)	13.6 (0.5)	13.0 (0.6)	0.6 (0.1)	0.6 (0.2)	-0.08 (0.36)	-0.19 (0.32)
	Non-dominant	0.33 (0.32)	0.52 (0.45)	13.3 (0.7)	12.8 (0.9)	0.6 (0.1)	0.6 (0.2)	-0.40 (0.45)	-0.57 (0.54)
	Intact	0.42 (0.67)	0.18 (0.47)	16.7 (2.4)	15.7 (2.9)	1.0 (0.3)	0.9 (0.3)	-0.50 (0.34)	-0.51 (0.74)
	Prosthetic	0.34 (0.41)	0.22 (0.48)	13.7 (1.4)	12.8 (1.2)	0.9 (0.3)	1.0 (0.3)	-0.25 (0.31)	-0.03 (0.36)
MLT	Dominant	-0.14 (0.31)	-0.14 (0.34)	13.3 (0.4)	12.7 (0.6)	0.8 (0.1)	0.9 (0.2)	0.14 (0.36)	0.18 (0.33)
	Non-dominant	-0.26 (0.42)	-0.20 (0.29)	12.6 (1.0)	12.0 (0.9)	0.8 (0.2)	0.7 (0.1)	0.27 (0.43)	0.24 (0.29)
	Intact	-0.32 (0.42)	-0.30 (0.43)	16.1 (2.3)	15.3 (2.7)	1.1 (0.2)	1.0 (0.3)	0.36 (0.40)	-0.07 (0.43)
	Prosthetic	-0.07 (0.39)	-0.08 (0.60)	13.5 (1.3)	12.8 (1.6)	1.0 (0.2)	1.0 (0.3)	-0.19 (0.30)	-0.14 (0.61)
HL	Dominant	0.07 (0.40)	0.22 (0.49)	13.5 (0.6)	13.3 (0.7)	1.4 (0.3)	1.5 (0.4)	-0.04 (0.36)	-0.33 (0.48)
	Non-dominant	**-0.20 (0.37)**	**0.30 (0.34)**	12.8 (0.9)	12.7 (1.2)	**1.3 (0.3)**	**1.5 (0.3)**	**0.17 (0.69)**	**-0.46 (0.52)**
	Intact	**0.11 (0.57)**	**0.76 (0.77)**	16.3 (2.4)	16.2 (2.8)	**1.6 (0.3)**	**2.0 (0.6)**	**-0.04 (0.37)**	**-0.67 (0.83)**
	Prosthetic	**0.03 (0.52)**	**0.20 (0.85)**	13.5 (1.6)	13.1 (1.7)	**1.8 (0.3)**	**2.3 (0.8)**	**-0.06 (0.65)**	**-0.36 (0.86)**
RO	Dominant	**-0.07 (0.38)**	**0.15 (0.21)**	13.4 (0.4)	13.0 (0.6)	**1.0 (0.2)**	**0.9 (0.2)**	**0.02 (0.36)**	**-0.11 (0.22)**
	Non-dominant	-0.19 (0.31)	-0.03 (0.30)	12.6 (0.8)	12.3 (0.9)	1.0 (0.2)	0.9 (0.2)	0.43 (0.53)	0.18 (0.26)
	Intact	-0.10 (0.68)	0.16 (0.61)	16.3 (2.1)	15.6 (2.6)	1.4 (0.4)	1.3 (0.4)	0.08 (0.81)	0.02 (0.61)
	Prosthetic	-0.04 (0.52)	0.13 (0.41)	13.4 (1.2)	12.8 (1.4)	1.3 (0.4)	1.4 (0.3)	0.20 (0.36)	0.03 (0.45)

Table A10. Mean and mean of standard deviations (mSD) for stance time.

Stance time (% stride time)

		During-Before		During				After-During	
		Fixed	Self-paced	Fixed	Self-paced	Fixed	Self-paced	Fixed	Self-paced
Activity	Limb	Mean	Mean	Mean	Mean	mSD	mSD	Mean	Mean
DS	Dominant	-0.96 (0.53)	-1.18 (0.61)	62.0 (0.9)	61.5 (0.9)	0.8 (0.1)	0.9 (0.2)	1.05 (0.51)	1.13 (0.67)
	Non-dominant	-1.30 (0.76)	-1.32 (0.75)	61.9 (0.7)	61.4 (0.9)	0.7 (0.1)	0.9 (0.2)	1.28 (0.68)	1.41 (0.74)
	Intact	-0.88 (1.29)	-0.61 (1.33)	64.8 (2.6)	64.5 (3.0)	1.5 (0.7)	1.5 (0.5)	0.97 (1.25)	0.52 (1.12)
	Prosthetic	-1.35 (1.78)	-1.29 (1.36)	62.4 (1.6)	62.2 (1.4)	1.4 (0.9)	1.5 (0.3)	1.31 (1.60)	1.13 (1.53)
US	Dominant	**0.37 (0.73)**	**0.81 (0.77)**	63.5 (0.8)	63.4 (0.8)	0.8 (0.2)	0.9 (0.3)	-0.38 (0.77)	-0.73 (0.77)
	Non-dominant	**0.40 (0.66)**	**1.02 (0.45)**	63.7 (0.6)	63.8 (0.7)	0.8 (0.2)	0.9 (0.3)	-0.52 (0.62)	-1.01 (0.54)
	Intact	**0.06 (1.49)**	**0.59 (1.37)**	66.0 (2.7)	65.5 (2.9)	1.2 (0.4)	1.3 (0.4)	-0.33 (1.56)	-0.49 (1.38)
	Prosthetic	**-0.71 (0.78)**	**-0.46 (1.17)**	63.3 (1.8)	62.8 (2.1)	1.2 (0.6)	1.2 (0.2)	0.87 (0.62)	0.55 (1.09)
TS	Dominant	0.02 (0.33)	0.08 (0.35)	63.1 (0.7)	62.6 (0.8)	0.7 (0.2)	0.6 (0.1)	-0.06 (0.33)	-0.12 (0.33)
	Non-dominant	-0.08 (0.40)	0.06 (0.63)	63.1 (0.6)	62.6 (0.7)	0.7 (0.1)	0.6 (0.1)	0.16 (0.39)	-0.02 (0.58)
	Intact	-0.12 (0.40)	0.06 (0.48)	65.7 (2.1)	64.7 (2.5)	1.0 (0.4)	1.0 (0.3)	0.07 (0.37)	0.18 (0.53)
	Prosthetic	0.10 (0.43)	0.06 (0.23)	64.0 (1.6)	63.3 (1.9)	1.1 (0.4)	1.0 (0.2)	0.05 (0.38)	-0.16 (0.31)
BS	Dominant	-0.18 (0.30)	-0.22 (0.23)	63.0 (0.7)	62.4 (0.6)	0.7 (0.2)	0.7 (0.1)	0.19 (0.34)	0.13 (0.35)
	Non-dominant	-0.02 (0.23)	0.03 (0.48)	63.2 (0.5)	62.7 (0.7)	0.7 (0.2)	0.7 (0.1)	-0.08 (0.50)	-0.12 (0.64)
	Intact	-0.02 (0.51)	0.02 (0.72)	65.8 (2.5)	65.0 (2.9)	1.0 (0.3)	1.0 (0.3)	-0.12 (0.36)	-0.32 (0.76)
	Prosthetic	0.04 (0.33)	-0.08 (0.45)	63.9 (1.6)	63.1 (1.7)	0.9 (0.5)	1.0 (0.2)	0.03 (0.39)	0.13 (0.25)
MLT	Dominant	-0.16 (0.22)	-0.28 (0.30)	63.0 (0.7)	62.3 (0.8)	1.2 (0.4)	1.2 (0.3)	0.09 (0.29)	0.25 (0.29)
	Non-dominant	-0.26 (0.39)	-0.11 (0.38)	62.9 (0.7)	62.5 (0.7)	1.1 (0.4)	0.9 (0.3)	0.32 (0.41)	0.18 (0.30)
	Intact	0.13 (0.30)	-0.11 (0.35)	65.9 (2.3)	65.0 (2.5)	1.5 (0.3)	1.4 (0.3)	-0.07 (0.38)	-0.25 (0.30)
	Prosthetic	-0.49 (0.30)	-0.28 (0.64)	63.7 (1.5)	63.0 (2.1)	1.2 (0.3)	1.1 (0.4)	0.22 (0.27)	0.01 (0.51)
HL	Dominant	**0.00 (0.26)**	**0.27 (0.39)**	63.1 (0.7)	63.0 (0.8)	**1.1 (0.3)**	**1.3 (0.2)**	**0.01 (0.36)**	**-0.38 (0.29)**
	Non-dominant	**-0.17 (0.43)**	**0.19 (0.44)**	63.1 (0.7)	63.0 (1.0)	**1.1 (0.3)**	**1.2 (0.4)**	**0.18 (0.66)**	**-0.30 (0.54)**
	Intact	**0.12 (0.49)**	**0.66 (0.81)**	65.8 (2.4)	65.6 (2.5)	**1.5 (0.4)**	**2.0 (0.9)**	**-0.03 (0.37)**	**-0.68 (0.90)**
	Prosthetic	**-0.11 (0.30)**	**0.21 (0.68)**	63.9 (1.5)	63.5 (1.5)	**1.5 (0.5)**	**2.1 (1.3)**	**0.03 (0.49)**	**-0.22 (0.63)**
RO	Dominant	-0.11 (0.32)	0.13 (0.14)	63.0 (0.6)	62.7 (0.6)	1.1 (0.2)	1.0 (0.2)	0.14 (0.40)	-0.04 (0.23)
	Non-dominant	-0.16 (0.44)	-0.06 (0.38)	63.0 (0.6)	62.6 (0.8)	1.0 (0.2)	0.9 (0.2)	0.30 (0.59)	0.14 (0.31)
	Intact	0.12 (0.53)	0.19 (0.62)	66.0 (2.2)	65.2 (2.8)	1.3 (0.4)	1.3 (0.3)	-0.12 (0.46)	-0.10 (0.66)
	Prosthetic	-0.29 (0.51)	-0.01 (0.56)	63.7 (1.5)	63.1 (1.6)	1.4 (0.6)	1.2 (0.3)	0.43 (0.58)	0.23 (0.52)

Table A11. Mean and mean of standard deviations (mSD) for swing time.

Swing time (% stride time)

		During-Before		During				After-During	
		Fixed	Self-paced	Fixed	Self-paced	Fixed	Self-paced	Fixed	Self-paced
Activity	Limb	Mean	Mean	Mean	Mean	mSD	mSD	Mean	Mean
DS	Dominant	0.96 (0.53)	1.18 (0.61)	38.0 (0.9)	38.5 (0.9)	0.8 (0.1)	0.9 (0.2)	-1.05 (0.51)	-1.13 (0.67)
	Non-dominant	1.30 (0.76)	1.32 (0.75)	38.1 (0.7)	38.6 (0.9)	0.7 (0.1)	0.9 (0.2)	-1.28 (0.68)	-1.41 (0.74)
	Intact	0.88 (1.29)	0.61 (1.33)	35.2 (2.6)	35.5 (3.0)	1.5 (0.7)	1.5 (0.5)	-0.97 (1.25)	-0.52 (1.12)
	Prosthetic	1.35 (1.78)	1.29 (1.36)	37.6 (1.6)	37.8 (1.4)	1.4 (0.9)	1.5 (0.3)	-1.31 (1.60)	-1.13 (1.53)
US	Dominant	**-0.37 (0.73)**	**-0.81 (0.77)**	36.5 (0.8)	36.6 (0.8)	0.8 (0.2)	0.9 (0.3)	0.38 (0.77)	0.73 (0.77)
	Non-dominant	**-0.40 (0.66)**	**-1.02 (0.45)**	36.3 (0.6)	36.2 (0.7)	0.8 (0.2)	0.9 (0.3)	0.52 (0.62)	1.01 (0.54)
	Intact	**-0.06 (1.49)**	**-0.59 (1.37)**	34.0 (2.7)	34.5 (2.9)	1.2 (0.4)	1.3 (0.4)	0.33 (1.56)	0.49 (1.38)
	Prosthetic	**0.71 (0.78)**	**0.46 (1.17)**	36.7 (1.8)	37.2 (2.1)	1.2 (0.6)	1.2 (0.2)	-0.87 (0.62)	-0.55 (1.09)
TS	Dominant	-0.02 (0.33)	-0.08 (0.35)	36.9 (0.7)	37.4 (0.8)	0.7 (0.2)	0.6 (0.1)	0.06 (0.33)	0.12 (0.33)
	Non-dominant	0.08 (0.40)	-0.06 (0.63)	36.9 (0.6)	37.4 (0.7)	0.7 (0.1)	0.6 (0.1)	-0.16 (0.39)	0.02 (0.58)
	Intact	0.12 (0.40)	-0.06 (0.48)	34.3 (2.1)	35.3 (2.5)	1.0 (0.4)	1.0 (0.3)	-0.07 (0.37)	-0.18 (0.53)
	Prosthetic	-0.10 (0.43)	-0.06 (0.23)	36.0 (1.6)	36.7 (1.9)	1.1 (0.4)	1.0 (0.2)	-0.05 (0.38)	0.16 (0.31)
BS	Dominant	0.18 (0.30)	0.22 (0.23)	37.0 (0.7)	37.6 (0.6)	0.7 (0.2)	0.7 (0.1)	-0.19 (0.34)	-0.13 (0.35)
	Non-dominant	0.02 (0.23)	-0.03 (0.48)	36.8 (0.5)	37.3 (0.7)	0.7 (0.2)	0.7 (0.1)	0.08 (0.50)	0.12 (0.64)
	Intact	0.02 (0.51)	-0.02 (0.72)	34.2 (2.5)	35.0 (2.9)	1.0 (0.3)	1.0 (0.3)	0.12 (0.36)	0.32 (0.76)
	Prosthetic	-0.04 (0.33)	0.08 (0.45)	36.1 (1.6)	36.9 (1.7)	0.9 (0.5)	1.0 (0.2)	-0.03 (0.39)	-0.13 (0.25)
MLT	Dominant	0.16 (0.22)	0.28 (0.30)	37.0 (0.7)	37.7 (0.8)	1.2 (0.4)	1.2 (0.3)	-0.09 (0.29)	-0.25 (0.29)
	Non-dominant	0.26 (0.39)	0.11 (0.38)	37.1 (0.7)	37.5 (0.7)	1.1 (0.4)	0.9 (0.3)	-0.32 (0.41)	-0.18 (0.30)
	Intact	-0.13 (0.30)	0.11 (0.35)	34.1 (2.3)	35.0 (2.5)	1.5 (0.3)	1.4 (0.3)	0.07 (0.38)	0.25 (0.30)
	Prosthetic	0.49 (0.30)	0.28 (0.64)	36.3 (1.5)	37.0 (2.1)	1.2 (0.3)	1.1 (0.4)	-0.22 (0.27)	-0.01 (0.51)
HL	Dominant	**0.00 (0.26)**	**-0.27 (0.39)**	36.9 (0.7)	37.0 (0.8)	**1.1 (0.3)**	**1.3 (0.2)**	**-0.01 (0.36)**	**0.38 (0.29)**
	Non-dominant	**0.17 (0.43)**	**-0.19 (0.44)**	36.9 (0.7)	37.0 (1.0)	**1.1 (0.3)**	**1.2 (0.4)**	**-0.18 (0.66)**	**0.30 (0.54)**
	Intact	**-0.12 (0.49)**	**-0.66 (0.81)**	34.2 (2.4)	34.4 (2.5)	**1.5 (0.4)**	**2.0 (0.9)**	**0.03 (0.37)**	**0.68 (0.90)**
	Prosthetic	**0.11 (0.30)**	**-0.21 (0.68)**	36.1 (1.5)	36.5 (1.5)	**1.5 (0.5)**	**2.1 (1.3)**	**-0.03 (0.49)**	**0.22 (0.63)**
RO	Dominant	0.11 (0.32)	-0.13 (0.14)	37.0 (0.6)	37.3 (0.6)	1.1 (0.2)	1.0 (0.2)	-0.14 (0.40)	0.04 (0.23)
	Non-dominant	0.16 (0.44)	0.06 (0.38)	37.0 (0.6)	37.4 (0.8)	1.0 (0.2)	0.9 (0.2)	-0.30 (0.59)	-0.14 (0.31)
	Intact	-0.12 (0.53)	-0.19 (0.62)	34.0 (2.2)	34.8 (2.8)	1.3 (0.4)	1.3 (0.3)	0.12 (0.46)	0.10 (0.66)
	Prosthetic	0.29 (0.51)	0.01 (0.56)	36.3 (1.5)	36.9 (1.6)	1.4 (0.6)	1.2 (0.3)	-0.43 (0.58)	-0.23 (0.52)

Table A12. Able-bodied range-of-motion for the ankle, knee, hip, pelvis, and trunk. Standard deviation is in brackets. Main effects for speed condition were observed for hip and pelvis (boldfaced parameters). However, significant differences between fixed and self-paced speed conditions were not significant with a treadmill speed covariate.

		Fixed Speed							
Able-bodied		LW	DS	US	TS	BS	MLT	HL	RO
Dominant Limb	Ankle dorsi/plantar flex	30.7 (2.3)	29.0 (1.4)	37.0 (3.5)	31.4 (2.4)	31.5 (2.5)	30.0 (2.5)	30.6 (2.5)	30.4 (2.8)
	Knee flex/ext	71.7 (1.1)	76.8 (1.6)	61.4 (2.7)	71.8 (1.5)	70.4 (1.0)	70.3 (1.2)	70.2 (1.2)	68.3 (1.8)
	Hip flex/ext	37.8 (2.3)	29.0 (2.2)	55.6 (2.6)	39.6 (2.1)	37.3 (2.1)	38.3 (2.0)	39.9 (2.2)	40.3 (1.9)
	Hip ad/abduction	13.2 (2.0)	12.5 (1.9)	13.1 (1.8)	13.7 (1.7)	12.8 (1.8)	13.1 (1.7)	12.9 (1.6)	12.6 (1.8)
	Pelvis tilt	4.4 (0.7)	4.9 (0.9)	4.3 (0.8)	4.5 (0.7)	4.5 (0.6)	4.6 (0.5)	4.9 (0.6)	5.1 (0.6)
	Pelvis obliquity	6.6 (0.8)	6.2 (0.9)	6.9 (1.4)	6.7 (0.9)	6.6 (0.8)	6.5 (0.9)	6.6 (0.8)	6.2 (0.8)
	Pelvis rotation	8.8 (1.6)	8.4 (1.6)	7.1 (1.3)	8.5 (1.7)	8.7 (1.5)	9.0 (1.1)	8.6 (1.5)	8.8 (1.2)
	Trunk flex/ext	3.4 (0.3)	3.0 (0.4)	4.3 (0.3)	3.7 (0.3)	3.5 (0.4)	3.5 (0.4)	4.3 (0.4)	4.1 (0.3)
	Trunk lateral flex/ext	3.2 (0.6)	2.6 (0.3)	4.6 (0.9)	3.3 (0.5)	3.3 (0.5)	3.8 (0.4)	3.4 (0.4)	4.1 (0.5)
	Trunk rotation	6.8 (0.6)	6.0 (0.8)	7.7 (1.8)	6.7 (0.6)	6.7 (0.7)	7.6 (0.9)	7.0 (0.6)	8.0 (1.2)
Non-dominant Limb	Ankle dorsi/plantar flex	29.7 (2.2)	28.5 (2.3)	35.1 (2.4)	30.5 (2.1)	30.6 (2.5)	29.4 (2.0)	29.6 (2.2)	29.2 (2.2)
	Knee flex/ext	71.4 (1.4)	77.8 (1.2)	61.9 (3.1)	71.4 (1.7)	70.4 (1.0)	70.5 (1.1)	70.4 (1.5)	69.0 (1.6)
	Hip flex/ext	39.7 (2.3)	30.2 (2.3)	57.0 (2.8)	41.2 (2.6)	39.3 (2.5)	40.1 (2.2)	42.2 (2.4)	41.9 (2.2)
	Hip ad/abduction	14.1 (2.1)	13.1 (2.2)	13.7 (1.4)	14.6 (1.8)	13.6 (1.8)	14.1 (1.6)	13.6 (1.7)	13.5 (1.8)
	Pelvis tilt	4.4 (0.7)	5.0 (0.9)	4.3 (0.8)	4.6 (0.6)	4.5 (0.7)	4.6 (0.5)	5.0 (0.7)	5.1 (0.7)
	Pelvis obliquity	6.6 (0.8)	6.2 (0.9)	6.9 (1.4)	6.7 (0.8)	6.7 (0.9)	6.5 (0.9)	6.6 (0.8)	6.3 (0.8)
	Pelvis rotation	8.8 (1.6)	8.5 (1.5)	7.2 (1.3)	8.7 (1.5)	8.5 (1.7)	8.9 (1.1)	8.5 (1.4)	8.7 (1.2)
	Trunk flex/ext	3.3 (0.3)	3.1 (0.4)	4.2 (0.3)	3.8 (0.4)	3.5 (0.3)	3.5 (0.3)	4.5 (0.5)	4.0 (0.3)
	Trunk lateral flex/ext	3.2 (0.6)	2.5 (0.3)	4.6 (0.9)	3.2 (0.5)	3.3 (0.5)	3.7 (0.5)	3.4 (0.4)	4.1 (0.4)
	Trunk rotation	6.9 (0.7)	6.1 (0.7)	7.8 (1.8)	6.8 (0.8)	6.7 (0.6)	7.6 (1.0)	7.1 (0.7)	8.1 (1.3)
		Self-paced Speed							
Able-bodied		LW	DS	US	TS	BS	MLT	HL	RO
Dominant Limb	Ankle dorsi/plantar flex	31.6 (4.9)	28.6 (3.5)	36.2 (6.4)	32.8 (4.7)	32.4 (4.7)	30.6 (4.1)	30.9 (4.7)	31.2 (4.7)
	Knee flex/ext	71.2 (1.5)	77.0 (3.3)	61.3 (4.8)	71.3 (2.4)	70.1 (1.7)	70.0 (1.6)	69.3 (2.1)	68.4 (2.3)
	Hip flex/ext	39.9 (3.6)	33.0 (5.0)	56.2 (4.4)	42.4 (3.4)	40.6 (3.5)	41.1 (3.2)	41.7 (4.5)	43.1 (3.2)
	Hip ad/abduction	13.7 (3.5)	12.9 (3.1)	12.8 (2.9)	14.4 (3.4)	13.4 (3.2)	13.7 (2.8)	12.9 (2.9)	13.1 (3.2)
	Pelvis tilt	4.5 (1.1)	4.9 (1.4)	4.2 (1.4)	4.6 (1.2)	4.7 (1.1)	4.9 (1.1)	5.1 (1.0)	5.2 (1.1)
	Pelvis obliquity	7.1 (1.6)	6.5 (1.3)	6.7 (2.5)	7.3 (2.0)	7.3 (1.8)	7.1 (1.9)	6.6 (1.7)	6.8 (1.7)
	Pelvis rotation	9.8 (3.4)	9.7 (3.2)	7.9 (2.6)	10.0 (3.2)	10.2 (3.3)	10.1 (2.5)	9.0 (2.2)	9.4 (2.5)
	Trunk flex/ext	6.6 (2.9)	5.7 (3.3)	13.4 (4.5)	7.6 (3.3)	7.5 (3.6)	8.3 (3.4)	8.7 (3.6)	8.8 (3.5)
	Trunk lateral flex/ext	3.5 (0.5)	3.3 (0.7)	4.2 (0.7)	3.9 (0.6)	3.6 (0.5)	3.8 (0.5)	4.6 (0.7)	4.2 (0.6)
	Trunk rotation	3.3 (0.9)	2.8 (0.7)	4.8 (1.8)	3.6 (0.9)	3.4 (1.0)	4.0 (0.7)	3.8 (0.7)	4.3 (0.8)
Non-dominant Limb	Ankle dorsi/plantar flex	30.4 (3.7)	27.8 (4.2)	34.9 (4.1)	31.0 (3.1)	31.2 (4.1)	29.9 (3.3)	29.4 (3.5)	29.7 (3.6)
	Knee flex/ext	71.5 (2.0)	78.4 (2.7)	62.6 (5.0)	71.9 (3.4)	70.4 (2.3)	70.7 (2.3)	70.2 (3.3)	70.0 (2.9)
	Hip flex/ext	41.0 (4.2)	33.2 (5.6)	57.5 (5.5)	44.1 (5.1)	41.8 (5.0)	42.8 (4.5)	43.3 (4.7)	44.7 (4.6)
	Hip ad/abduction	14.4 (3.6)	13.5 (3.7)	13.3 (2.5)	15.3 (3.4)	14.3 (3.6)	14.5 (2.9)	13.8 (3.0)	14.0 (3.3)
	Pelvis tilt	4.6 (1.1)	5.0 (1.5)	4.3 (1.4)	4.7 (1.1)	4.7 (1.1)	4.8 (1.0)	5.2 (1.1)	5.1 (1.1)
	Pelvis obliquity	7.1 (1.6)	6.5 (1.3)	6.7 (2.5)	7.3 (1.8)	7.3 (2.0)	7.1 (1.9)	6.6 (1.7)	6.8 (1.7)
	Pelvis rotation	9.8 (3.4)	9.7 (3.1)	8.0 (2.6)	10.2 (3.4)	9.9 (3.2)	10.0 (2.5)	8.9 (2.2)	9.3 (2.4)
	Trunk flex/ext	6.6 (2.9)	5.7 (3.3)	13.3 (4.6)	7.8 (3.6)	7.4 (3.3)	8.3 (3.4)	8.6 (3.7)	8.8 (3.5)
	Trunk lateral flex/ext	3.4 (0.5)	3.3 (0.7)	4.2 (0.6)	3.8 (0.7)	3.8 (0.6)	3.7 (0.5)	4.6 (0.6)	4.1 (0.5)
	Trunk rotation	3.3 (0.9)	2.9 (0.7)	4.7 (1.8)	3.4 (1.0)	3.6 (0.8)	4.1 (0.8)	3.8 (0.7)	4.4 (0.8)

Table A13. Transtibial amputee range-of-motion for the ankle, knee, hip, pelvis, and trunk. Standard deviation is in brackets. Main effects for speed condition were observed for hip and pelvis (boldfaced parameters). However, significant differences between fixed and self-paced speed conditions were not significant with a treadmill speed covariate.

		Fixed Speed							
Transtibial Amputee		LW	DS	US	TS	BS	MLT	HL	RO
Intact Limb	Ankle dorsi/plantar flex	30.2 (2.3)	31.3 (3.8)	39.8 (3.4)	31.1 (2.1)	30.4 (2.5)	29.3 (2.3)	30.1 (2.3)	29.4 (2.3)
	Knee flex/ext	66.7 (2.5)	72.9 (3.1)	55.7 (2.2)	66.9 (2.3)	64.7 (2.6)	65.6 (2.3)	66.4 (2.1)	65.6 (2.2)
	Hip flex/ext	38.3 (3.5)	27.8 (2.6)	55.3 (3.8)	41.2 (3.6)	39.5 (3.5)	39.4 (3.4)	40.5 (3.8)	41.3 (3.6)
	Hip ad/abduction	10.9 (1.7)	11.1 (3.2)	13.8 (1.6)	12.0 (1.5)	10.4 (1.6)	11.4 (1.6)	10.9 (1.5)	11.1 (1.6)
	Pelvis tilt	3.9 (0.6)	5.5 (0.8)	5.0 (1.0)	4.1 (0.8)	4.4 (0.8)	4.2 (0.6)	4.5 (0.6)	4.7 (0.7)
	Pelvis obliquity	5.3 (0.9)	8.4 (1.5)	9.3 (0.9)	5.3 (0.9)	5.1 (0.8)	5.2 (0.7)	5.5 (0.8)	5.8 (0.8)
	Pelvis rotation	10.1 (1.7)	11.0 (2.3)	8.2 (1.4)	10.3 (1.5)	10.2 (1.5)	10.0 (1.4)	10.1 (1.4)	10.6 (1.7)
	Trunk flex/ext	3.7 (0.6)	4.5 (0.8)	4.5 (0.9)	3.9 (0.6)	4.3 (1.1)	3.9 (0.6)	4.6 (0.5)	4.4 (0.6)
	Trunk lateral flex/ext	6.0 (0.9)	5.8 (1.0)	9.4 (1.5)	6.7 (0.9)	6.4 (0.9)	6.8 (1.0)	6.9 (1.1)	7.5 (0.9)
	Trunk rotation	9.2 (1.4)	8.5 (1.7)	13.0 (1.6)	9.5 (1.8)	9.5 (1.8)	9.8 (1.6)	9.6 (1.4)	11.1 (1.7)
Prosthetic Limb	Ankle dorsi/plantar flex	19.4 (2.3)	19.8 (2.2)	17.4 (2.2)	19.6 (2.4)	19.0 (2.2)	19.3 (2.2)	19.2 (2.2)	19.3 (2.2)
	Knee flex/ext	63.2 (3.9)	71.5 (3.4)	60.9 (5.9)	64.9 (4.2)	61.4 (3.8)	63.0 (3.8)	64.0 (4.2)	63.7 (4.2)
	Hip flex/ext	41.8 (2.9)	28.6 (3.0)	57.2 (4.5)	43.2 (2.7)	42.0 (3.3)	41.8 (2.5)	43.0 (2.9)	43.7 (2.9)
	Hip ad/abduction	10.0 (1.9)	11.8 (2.2)	13.9 (1.5)	11.3 (1.8)	9.8 (1.9)	10.7 (1.6)	10.3 (1.7)	10.5 (1.7)
	Pelvis tilt	4.0 (0.6)	5.4 (0.7)	4.9 (1.0)	4.4 (0.8)	4.0 (0.7)	4.2 (0.7)	4.5 (0.6)	4.5 (0.6)
	Pelvis obliquity	5.3 (0.9)	8.5 (1.5)	9.3 (1.0)	5.1 (0.8)	5.3 (0.9)	5.2 (0.7)	5.6 (0.7)	5.8 (0.8)
	Pelvis rotation	10.1 (1.7)	11.0 (2.3)	8.3 (1.4)	10.1 (1.4)	10.3 (1.5)	10.0 (1.4)	10.3 (1.4)	10.3 (1.7)
	Trunk flex/ext	3.8 (0.6)	4.6 (0.8)	4.5 (1.1)	4.5 (1.1)	3.8 (0.6)	4.0 (0.7)	4.9 (0.6)	4.4 (0.7)
	Trunk lateral flex/ext	6.0 (0.9)	6.0 (1.0)	9.3 (1.4)	6.5 (1.0)	6.7 (1.0)	6.7 (1.0)	7.0 (1.1)	7.3 (0.9)
	Trunk rotation	9.1 (1.4)	8.5 (1.7)	13.0 (1.6)	9.4 (1.8)	9.5 (1.8)	9.8 (1.7)	9.6 (1.4)	10.9 (1.7)
		Self-paced Speed							
Transtibial Amputee		LW	DS	US	TS	BS	MLT	HL	RO
Intact Limb	Ankle dorsi/plantar flex	30.0 (4.9)	30.2 (6.6)	38.5 (5.7)	31.2 (4.8)	30.9 (4.4)	30.1 (4.9)	29.3 (4.9)	29.4 (4.9)
	Knee flex/ext	65.7 (4.7)	71.6 (6.3)	56.0 (4.1)	66.3 (5.0)	64.2 (5.6)	65.4 (4.9)	65.2 (4.3)	64.7 (5.1)
	Hip flex/ext	40.8 (6.1)	28.3 (5.6)	56.3 (7.3)	44.5 (7.1)	43.0 (6.4)	42.9 (6.8)	43.0 (7.9)	44.2 (6.4)
	Hip ad/abduction	11.5 (3.0)	11.3 (4.9)	14.1 (3.0)	12.9 (2.8)	11.4 (2.7)	12.5 (2.6)	11.5 (2.8)	12.1 (2.9)
	Pelvis tilt	3.8 (1.1)	5.6 (1.4)	4.8 (2.4)	4.2 (1.3)	4.3 (1.1)	4.2 (1.2)	4.7 (1.2)	4.7 (1.3)
	Pelvis obliquity	5.8 (2.5)	8.7 (2.9)	9.8 (2.6)	6.1 (2.6)	5.7 (2.3)	6.0 (2.3)	6.2 (2.1)	6.3 (2.3)
	Pelvis rotation	10.9 (3.6)	11.8 (6.0)	8.7 (2.4)	11.9 (3.4)	12.1 (3.8)	11.6 (4.0)	11.0 (3.6)	11.2 (3.6)
	Trunk flex/ext	12.8 (3.0)	12.9 (3.3)	22.9 (2.8)	14.7 (2.6)	14.0 (2.9)	14.2 (2.7)	14.6 (2.7)	15.3 (2.8)
	Trunk lateral flex/ext	3.8 (0.9)	4.4 (1.3)	4.8 (1.9)	4.3 (1.2)	4.0 (1.2)	4.1 (0.9)	4.8 (0.9)	4.5 (1.0)
	Trunk rotation	5.8 (1.6)	5.8 (1.4)	9.6 (2.4)	7.0 (1.9)	6.6 (1.9)	7.0 (1.6)	7.0 (2.1)	7.6 (1.8)
Prosthetic Limb	Ankle dorsi/plantar flex	19.8 (4.3)	19.7 (3.9)	17.6 (5.1)	20.6 (4.6)	20.0 (4.1)	20.3 (4.4)	19.3 (4.4)	20.1 (4.5)
	Knee flex/ext	63.9 (9.9)	72.1 (9.4)	62.1 (12.2)	66.9 (9.8)	63.7 (8.6)	65.1 (8.8)	64.5 (9.3)	65.5 (9.6)
	Hip flex/ext	42.4 (4.9)	28.2 (5.9)	57.9 (7.9)	45.7 (4.7)	44.3 (5.6)	44.2 (4.7)	44.0 (6.3)	45.5 (4.6)
	Hip ad/abduction	10.9 (3.8)	11.9 (4.6)	14.6 (3.0)	12.1 (3.6)	11.0 (3.7)	11.5 (3.2)	11.1 (3.6)	11.4 (3.6)
	Pelvis tilt	3.8 (1.1)	5.4 (1.3)	4.6 (2.2)	4.3 (1.1)	4.1 (1.3)	4.3 (1.1)	4.8 (1.3)	4.6 (1.2)
	Pelvis obliquity	5.7 (2.5)	8.7 (2.8)	9.7 (2.6)	5.7 (2.3)	6.0 (2.6)	6.0 (2.3)	6.3 (2.0)	6.3 (2.4)
	Pelvis rotation	10.9 (3.6)	12.0 (6.1)	9.0 (2.6)	12.1 (3.7)	11.8 (3.3)	11.7 (4.3)	10.9 (3.7)	11.1 (3.7)
	Trunk flex/ext	12.8 (3.1)	12.9 (3.3)	22.9 (2.8)	14.2 (3.0)	14.5 (2.6)	14.2 (2.8)	14.6 (2.9)	15.2 (2.9)
	Trunk lateral flex/ext	3.8 (0.9)	4.4 (1.2)	4.8 (1.9)	4.2 (1.2)	4.2 (1.0)	4.1 (0.9)	4.9 (1.0)	4.4 (1.1)
	Trunk rotation	5.9 (1.7)	5.9 (1.5)	9.4 (2.3)	6.7 (2.0)	7.1 (1.9)	6.9 (1.8)	7.1 (2.2)	7.7 (1.8)

Table A14. Able-bodied range-of-motion mean of the standard deviations (mSD) for the ankle, knee, hip, pelvis, and trunk. Standard deviation is in brackets. Kinematic variability was generally larger for self-paced compared to the fixed speed condition (boldfaced parameters). Differences between speed conditions for pelvis and trunk were also no longer significant with a speed mSD covariate, however, ankle and knee mSD during the rocky (RO) activity were significantly smaller for the self-paced speed condition than the fixed speed condition.

		Fixed Speed							
Able-bodied		LW	DS	US	TS	BS	MLT	HL	RO
Dominant Limb	Ankle dorsi/plantar flex	2.1 (0.4)	2.1 (0.3)	2.5 (0.4)	2.2 (0.5)	2.0 (0.4)	3.3 (0.6)	3.2 (0.5)	3.8 (0.6)
	Knee flex/ext	1.4 (0.3)	1.6 (0.2)	2.0 (0.3)	1.9 (0.5)	1.4 (0.3)	2.0 (0.4)	2.8 (0.4)	3.2 (0.8)
	Hip flex/ext	1.1 (0.1)	1.8 (0.3)	1.8 (0.2)	1.3 (0.3)	1.3 (0.2)	1.5 (0.3)	3.1 (0.4)	1.9 (0.2)
	Hip ad/abduction	0.9 (0.1)	1.2 (0.2)	1.0 (0.1)	0.9 (0.2)	0.9 (0.1)	1.3 (0.2)	1.2 (0.1)	1.5 (0.2)
	Pelvis tilt	0.6 (0.1)	0.7 (0.2)	0.8 (0.1)	0.7 (0.1)	0.7 (0.1)	0.8 (0.1)	0.8 (0.1)	1.0 (0.1)
	Pelvis obliquity	0.7 (0.1)	0.7 (0.1)	0.7 (0.1)	0.7 (0.1)	0.7 (0.1)	0.9 (0.2)	0.8 (0.1)	0.9 (0.1)
	Pelvis rotation	1.3 (0.2)	1.4 (0.3)	1.4 (0.1)	1.3 (0.2)	1.3 (0.1)	1.9 (0.4)	1.8 (0.3)	1.8 (0.2)
	Trunk flex/ext	0.6 (0.1)	0.6 (0.1)	0.7 (0.1)	0.6 (0.1)	0.6 (0.1)	0.6 (0.1)	0.6 (0.1)	0.6 (0.1)
	Trunk lateral flex/ext	0.7 (0.2)	0.7 (0.1)	1.0 (0.2)	0.8 (0.1)	0.8 (0.1)	1.1 (0.3)	0.9 (0.1)	1.3 (0.2)
	Trunk rotation	1.2 (0.1)	1.4 (0.3)	1.5 (0.3)	1.3 (0.2)	1.2 (0.2)	1.8 (0.4)	1.6 (0.3)	1.7 (0.3)
Non-dominant Limb	Ankle dorsi/plantar flex	1.7 (0.3)	1.9 (0.2)	2.2 (0.3)	1.8 (0.4)	1.7 (0.3)	2.5 (0.2)	2.7 (0.2)	2.8 (0.4)
	Knee flex/ext	1.2 (0.2)	1.6 (0.2)	2.1 (0.3)	1.9 (0.5)	1.3 (0.3)	1.9 (0.4)	3.0 (0.3)	3.1 (0.7)
	Hip flex/ext	1.1 (0.2)	1.7 (0.3)	1.7 (0.3)	1.3 (0.2)	1.3 (0.2)	1.6 (0.3)	3.1 (0.4)	2.2 (0.4)
	Hip ad/abduction	0.9 (0.2)	1.2 (0.2)	1.0 (0.1)	0.9 (0.1)	1.0 (0.2)	1.4 (0.3)	1.4 (0.1)	1.7 (0.3)
	Pelvis tilt	0.6 (0.1)	0.8 (0.2)	0.8 (0.1)	0.7 (0.1)	0.7 (0.2)	0.7 (0.2)	0.9 (0.1)	1.0 (0.1)
	Pelvis obliquity	0.7 (0.1)	0.7 (0.1)	0.7 (0.1)	0.6 (0.1)	0.7 (0.1)	0.9 (0.2)	0.8 (0.1)	0.9 (0.1)
	Pelvis rotation	1.3 (0.2)	1.5 (0.2)	1.5 (0.1)	1.4 (0.2)	1.3 (0.2)	2.0 (0.5)	1.8 (0.2)	1.7 (0.3)
	Trunk flex/ext	0.5 (0.1)	0.5 (0.1)	0.6 (0.1)	0.6 (0.1)	0.5 (0.1)	0.5 (0.1)	0.6 (0.1)	0.5 (0.1)
	Trunk lateral flex/ext	0.7 (0.2)	0.7 (0.1)	1.1 (0.2)	0.7 (0.1)	0.8 (0.2)	1.1 (0.3)	1.0 (0.2)	1.2 (0.2)
	Trunk rotation	1.2 (0.1)	1.4 (0.2)	1.5 (0.3)	1.3 (0.2)	1.3 (0.2)	1.7 (0.5)	1.6 (0.2)	1.6 (0.3)
		Self-paced Speed							
Able-bodied		LW	DS	US	TS	BS	MLT	HL	RO
Dominant Limb	Ankle dorsi/plantar flex	1.6 (0.3)	2.4 (0.5)	2.8 (0.7)	2.0 (0.4)	1.9 (0.3)	3.3 (1.0)	3.2 (1.1)	3.0 (0.8)
	Knee flex/ext	1.3 (0.5)	2.0 (0.6)	2.3 (0.9)	1.9 (0.8)	1.4 (0.7)	1.9 (0.7)	3.1 (1.0)	2.6 (0.6)
	Hip flex/ext	1.5 (0.7)	2.6 (1.0)	2.6 (0.9)	1.7 (0.5)	1.4 (0.5)	1.8 (0.6)	3.7 (1.0)	2.1 (0.7)
	Hip ad/abduction	0.9 (0.4)	1.1 (0.3)	1.2 (0.4)	0.9 (0.4)	1.1 (0.3)	1.6 (0.4)	1.4 (0.3)	1.4 (0.4)
	Pelvis tilt	0.6 (0.1)	0.6 (0.2)	0.7 (0.3)	0.6 (0.1)	0.7 (0.3)	0.7 (0.3)	0.6 (0.3)	0.7 (0.3)
	Pelvis obliquity	0.7 (0.3)	0.8 (0.2)	0.8 (0.5)	0.7 (0.3)	0.8 (0.3)	1.0 (0.4)	1.0 (0.3)	1.0 (0.4)
	Pelvis rotation	1.5 (0.4)	1.9 (0.6)	1.8 (0.8)	1.6 (0.4)	1.5 (0.4)	2.2 (0.5)	2.2 (0.6)	2.0 (0.4)
	Trunk flex/ext	0.7 (0.2)	1.0 (0.4)	1.3 (0.6)	0.8 (0.5)	1.0 (0.5)	1.0 (0.4)	1.5 (0.5)	1.0 (0.3)
	Trunk lateral flex/ext	0.6 (0.1)	0.7 (0.2)	0.7 (0.2)	0.6 (0.2)	0.7 (0.1)	0.7 (0.2)	1.2 (0.4)	1.0 (0.3)
	Trunk rotation	0.7 (0.2)	0.7 (0.1)	1.1 (0.5)	0.8 (0.3)	0.7 (0.2)	1.2 (0.5)	1.0 (0.2)	1.2 (0.4)
Non-dominant Limb	Ankle dorsi/plantar flex	1.6 (0.4)	2.1 (0.5)	2.4 (0.6)	1.8 (0.5)	1.8 (0.7)	2.6 (0.6)	2.9 (0.6)	2.7 (0.5)
	Knee flex/ext	1.3 (0.7)	1.9 (0.5)	2.4 (1.5)	1.7 (0.6)	1.5 (0.8)	1.8 (0.7)	3.0 (0.7)	2.5 (0.7)
	Hip flex/ext	1.5 (0.6)	2.3 (0.9)	2.3 (0.5)	1.4 (0.4)	1.6 (0.5)	2.0 (0.6)	3.4 (0.8)	2.1 (0.9)
	Hip ad/abduction	0.8 (0.2)	1.2 (0.4)	1.1 (0.4)	0.9 (0.2)	0.9 (0.3)	1.6 (0.4)	1.4 (0.1)	1.6 (0.4)
	Pelvis tilt	0.7 (0.2)	0.6 (0.2)	0.7 (0.3)	0.7 (0.2)	0.6 (0.2)	0.7 (0.3)	0.7 (0.3)	0.7 (0.3)
	Pelvis obliquity	0.7 (0.3)	0.8 (0.2)	0.9 (0.4)	0.8 (0.3)	0.7 (0.4)	1.0 (0.3)	0.9 (0.3)	1.1 (0.3)
	Pelvis rotation	1.5 (0.5)	2.0 (0.5)	1.9 (0.8)	1.5 (0.4)	1.5 (0.5)	2.0 (0.5)	2.1 (0.6)	2.0 (0.5)
	Trunk flex/ext	0.7 (0.3)	1.0 (0.4)	1.3 (0.6)	0.9 (0.4)	0.9 (0.5)	1.0 (0.3)	1.6 (0.5)	1.0 (0.3)
	Trunk lateral flex/ext	0.5 (0.1)	0.7 (0.2)	0.7 (0.2)	0.7 (0.2)	0.6 (0.2)	0.7 (0.2)	1.1 (0.4)	0.9 (0.3)
	Trunk rotation	0.7 (0.1)	0.7 (0.1)	1.1 (0.6)	0.7 (0.2)	0.8 (0.2)	1.2 (0.4)	1.1 (0.3)	1.3 (0.3)

Table A15. Transtibial amputee range-of-motion mean of the standard deviations (mSD) for the ankle, knee, hip, pelvis, and trunk. Standard deviation is in brackets. Kinematic variability was generally larger for self-paced compared to the fixed speed condition (boldfaced parameters). Differences between speed conditions were no longer significant with a speed mSD covariate.

		Fixed Speed							
Transtibial Amputee		LW	DS	US	TS	BS	MLT	HL	RO
Intact Limb	**Ankle dorsi/plantar flex**	2.5 (0.4)	3.6 (1.0)	2.8 (0.4)	2.9 (0.5)	2.6 (0.4)	3.5 (0.7)	3.8 (0.4)	3.7 (0.5)
	Knee flex/ext	1.6 (0.2)	2.1 (0.4)	3.0 (0.6)	2.1 (0.4)	1.5 (0.2)	1.8 (0.3)	2.8 (0.3)	2.3 (0.3)
	Hip flex/ext	1.6 (0.3)	2.3 (0.5)	2.1 (0.4)	1.7 (0.3)	1.7 (0.3)	1.9 (0.3)	2.5 (0.3)	2.3 (0.5)
	Hip ad/abduction	1.3 (0.2)	1.7 (0.4)	1.8 (0.4)	1.1 (0.2)	1.6 (0.4)	2.0 (0.3)	1.6 (0.2)	1.8 (0.3)
	Pelvis tilt	0.8 (0.1)	1.3 (0.3)	1.2 (0.3)	0.9 (0.2)	1.0 (0.2)	1.0 (0.2)	1.2 (0.2)	1.3 (0.2)
	Pelvis obliquity	0.8 (0.2)	1.2 (0.2)	1.2 (0.3)	0.7 (0.1)	0.8 (0.1)	1.0 (0.2)	1.0 (0.1)	1.1 (0.1)
	Pelvis rotation	1.8 (0.3)	2.3 (0.5)	2.1 (0.5)	2.0 (0.3)	2.2 (0.3)	2.7 (0.5)	2.6 (0.6)	2.8 (0.5)
	Trunk flex/ext	0.7 (0.1)	0.8 (0.2)	0.9 (0.2)	0.8 (0.2)	0.8 (0.2)	0.8 (0.1)	0.8 (0.2)	0.8 (0.1)
	Trunk lateral flex/ext	1.2 (0.2)	1.1 (0.1)	2.1 (0.5)	1.4 (0.3)	1.3 (0.2)	1.7 (0.3)	1.8 (0.4)	1.8 (0.3)
	Trunk rotation	1.7 (0.4)	1.9 (0.4)	2.4 (0.5)	1.8 (0.5)	1.8 (0.4)	2.2 (0.5)	2.2 (0.3)	2.6 (0.6)
Prosthetic Limb	**Ankle dorsi/plantar flex**	0.5 (0.1)	0.7 (0.2)	0.9 (0.2)	0.6 (0.1)	0.6 (0.1)	0.8 (0.1)	1.1 (0.1)	0.9 (0.1)
	Knee flex/ext	1.5 (0.2)	2.4 (0.8)	3.3 (1.2)	1.8 (0.4)	1.7 (0.5)	2.1 (0.3)	2.4 (0.4)	2.1 (0.4)
	Hip flex/ext	1.6 (0.2)	3.1 (0.8)	2.1 (0.3)	1.9 (0.4)	1.6 (0.3)	1.9 (0.3)	2.6 (0.3)	2.2 (0.4)
	Hip ad/abduction	1.1 (0.1)	1.6 (0.4)	1.6 (0.3)	1.2 (0.3)	1.1 (0.2)	1.7 (0.3)	1.6 (0.3)	1.5 (0.2)
	Pelvis tilt	0.8 (0.1)	1.3 (0.3)	1.2 (0.3)	1.0 (0.2)	0.9 (0.2)	1.1 (0.2)	1.2 (0.1)	1.2 (0.2)
	Pelvis obliquity	0.8 (0.2)	1.1 (0.2)	1.3 (0.3)	0.8 (0.1)	0.8 (0.1)	0.9 (0.2)	1.0 (0.1)	1.0 (0.2)
	Pelvis rotation	1.7 (0.4)	2.2 (0.5)	1.9 (0.4)	2.2 (0.4)	1.8 (0.3)	2.6 (0.5)	2.3 (0.5)	2.6 (0.5)
	Trunk flex/ext	0.8 (0.1)	0.8 (0.1)	0.9 (0.2)	0.8 (0.2)	0.9 (0.3)	0.8 (0.2)	0.8 (0.2)	0.8 (0.2)
	Trunk lateral flex/ext	1.2 (0.2)	1.2 (0.3)	2.0 (0.5)	1.3 (0.2)	1.3 (0.3)	1.8 (0.4)	1.8 (0.3)	1.8 (0.3)
	Trunk rotation	1.7 (0.4)	1.8 (0.4)	2.3 (0.5)	1.7 (0.5)	1.9 (0.5)	2.0 (0.6)	2.0 (0.4)	2.6 (0.7)
		Self-paced Speed							
Transtibial Amputee		LW	DS	US	TS	BS	MLT	HL	RO
Intact Limb	**Ankle dorsi/plantar flex**	2.9 (0.9)	3.2 (1.2)	3.4 (1.2)	3.2 (1.3)	2.6 (0.5)	3.8 (0.9)	4.2 (1.4)	3.5 (0.6)
	Knee flex/ext	1.7 (0.5)	2.5 (1.0)	3.4 (2.0)	2.2 (0.8)	1.7 (0.4)	2.0 (0.5)	3.1 (0.8)	2.6 (0.6)
	Hip flex/ext	2.2 (1.0)	3.4 (1.1)	3.6 (1.6)	2.4 (0.8)	2.4 (0.9)	2.7 (1.0)	4.1 (1.0)	3.3 (2.0)
	Hip ad/abduction	1.6 (0.7)	1.7 (0.6)	2.0 (0.7)	1.4 (0.5)	1.6 (0.6)	2.0 (0.6)	2.0 (0.5)	1.9 (0.6)
	Pelvis tilt	0.9 (0.3)	1.0 (0.3)	0.9 (0.4)	1.0 (0.3)	1.0 (0.3)	0.9 (0.3)	0.9 (0.4)	1.0 (0.4)
	Pelvis obliquity	0.8 (0.6)	1.4 (0.4)	1.3 (0.6)	0.8 (0.5)	1.0 (0.7)	1.2 (0.6)	1.3 (0.3)	1.3 (0.6)
	Pelvis rotation	2.1 (0.9)	2.5 (1.4)	2.3 (0.8)	2.3 (0.8)	2.4 (0.9)	3.2 (1.6)	2.7 (1.1)	2.7 (1.2)
	Trunk flex/ext	0.9 (0.6)	1.8 (0.6)	1.6 (0.4)	1.2 (0.5)	1.2 (0.5)	1.3 (0.5)	1.9 (0.2)	1.5 (0.9)
	Trunk lateral flex/ext	0.7 (0.2)	1.3 (0.5)	1.2 (0.6)	0.9 (0.3)	0.8 (0.2)	1.0 (0.3)	1.3 (0.3)	1.1 (0.3)
	Trunk rotation	1.2 (0.4)	1.2 (0.3)	2.3 (1.0)	1.4 (0.4)	1.4 (0.4)	1.7 (0.5)	1.8 (0.6)	1.9 (0.7)
Prosthetic Limb	**Ankle dorsi/plantar flex**	0.9 (0.3)	1.0 (0.2)	1.1 (0.3)	0.9 (0.4)	0.9 (0.4)	1.0 (0.4)	1.5 (0.4)	1.2 (0.3)
	Knee flex/ext	1.9 (0.8)	3.2 (0.8)	3.1 (1.6)	2.1 (1.0)	2.1 (1.1)	2.1 (0.7)	3.9 (1.5)	3.1 (1.3)
	Hip flex/ext	2.5 (1.6)	3.5 (1.1)	3.4 (1.5)	2.6 (1.1)	2.2 (1.1)	2.5 (0.8)	4.5 (1.2)	3.3 (1.9)
	Hip ad/abduction	1.3 (0.7)	1.8 (0.7)	1.8 (0.7)	1.4 (0.8)	1.4 (0.5)	1.8 (0.6)	1.8 (0.6)	1.9 (0.7)
	Pelvis tilt	0.8 (0.3)	1.0 (0.4)	0.9 (0.3)	1.0 (0.3)	0.9 (0.3)	0.9 (0.3)	1.0 (0.3)	1.0 (0.4)
	Pelvis obliquity	0.8 (0.5)	1.4 (0.4)	1.5 (0.6)	1.0 (0.6)	0.8 (0.4)	1.2 (0.5)	1.3 (0.5)	1.3 (0.7)
	Pelvis rotation	2.2 (0.8)	2.5 (1.3)	2.2 (1.0)	2.4 (1.0)	2.2 (0.9)	2.8 (1.2)	2.5 (0.9)	2.9 (1.0)
	Trunk flex/ext	0.9 (0.4)	1.7 (0.6)	1.7 (0.5)	1.1 (0.5)	1.2 (0.6)	1.2 (0.5)	2.0 (0.4)	1.5 (0.8)
	Trunk lateral flex/ext	0.8 (0.3)	1.2 (0.3)	1.2 (0.5)	0.9 (0.3)	0.9 (0.3)	1.0 (0.4)	1.5 (0.5)	1.1 (0.2)
	Trunk rotation	1.2 (0.4)	1.2 (0.4)	2.2 (1.1)	1.4 (0.5)	1.4 (0.5)	1.7 (0.5)	2.2 (0.6)	2.0 (0.7)

Table A16. Able-bodied peak ground reaction force, only for walking activities when the motion platform was stationary. Standard deviation is in brackets. Main effects for speed condition were observed for all parameters (boldfaced). However, significant differences between fixed and self-paced speed conditions were not significant with a treadmill speed covariate.

				Fixed Speed		
Able-bodied		Level	Downhill	Uphill	Top-cross-slope	Bottom-cross-slope
Dominant Limb	**Vertical GRF early stance**	1.19 (0.07)	1.43 (0.08)	1.12 (0.06)	1.07 (0.34)	1.23 (0.07)
	Vertical GRF mid stance	0.75 (0.04)	0.70 (0.04)	0.71 (0.06)	0.66 (0.21)	0.75 (0.03)
	Vertical GRF late stance	1.12 (0.05)	0.94 (0.06)	1.16 (0.08)	0.99 (0.32)	1.12 (0.05)
	ML GRF early stance	-0.04 (0.02)	-0.03 (0.02)	-0.03 (0.02)	-0.03 (0.03)	-0.04 (0.02)
	ML GRF stance	0.09 (0.05)	0.11 (0.05)	0.07 (0.04)	0.07 (0.05)	0.09 (0.05)
	AP braking GRF	-0.22 (0.03)	-0.24 (0.02)	-0.20 (0.03)	-0.20 (0.07)	-0.22 (0.03)
	AP propulsion GRF	0.22 (0.03)	0.22 (0.03)	0.20 (0.04)	0.19 (0.07)	0.22 (0.03)
Non-dominant Limb	**Vertical GRF early stance**	1.17 (0.08)	1.38 (0.09)	1.10 (0.06)	1.17 (0.08)	1.10 (0.36)
	Vertical GRF mid stance	0.74 (0.05)	0.71 (0.04)	0.71 (0.05)	0.72 (0.05)	0.67 (0.22)
	Vertical GRF late stance	1.10 (0.05)	0.89 (0.07)	1.15 (0.08)	1.07 (0.06)	1.01 (0.32)
	ML GRF early stance	-0.03 (0.02)	-0.03 (0.02)	-0.03 (0.02)	-0.03 (0.02)	-0.04 (0.03)
	ML GRF stance	0.08 (0.04)	0.10 (0.05)	0.07 (0.03)	0.08 (0.04)	0.08 (0.05)
	AP braking GRF	-0.19 (0.02)	-0.21 (0.03)	-0.18 (0.02)	-0.20 (0.02)	-0.18 (0.06)
	AP propulsion GRF	0.23 (0.03)	0.23 (0.02)	0.22 (0.03)	0.24 (0.02)	0.21 (0.07)
				Self-paced Speed		
Able-bodied		Level	Downhill	Uphill	Top-cross-slope	Bottom-cross-slope
Dominant Limb	**Vertical GRF early stance**	1.24 (0.07)	1.51 (0.14)	1.12 (0.10)	1.22 (0.10)	1.32 (0.12)
	Vertical GRF mid stance	0.70 (0.06)	0.66 (0.07)	0.70 (0.06)	0.66 (0.08)	0.67 (0.07)
	Vertical GRF late stance	1.14 (0.06)	0.93 (0.07)	1.17 (0.10)	1.12 (0.09)	1.17 (0.08)
	ML GRF early stance	-0.04 (0.03)	-0.03 (0.02)	-0.04 (0.02)	-0.04 (0.03)	-0.05 (0.03)
	ML GRF stance	0.09 (0.05)	0.12 (0.06)	0.07 (0.04)	0.08 (0.05)	0.10 (0.05)
	AP braking GRF	-0.24 (0.03)	-0.27 (0.05)	-0.20 (0.04)	-0.25 (0.04)	-0.27 (0.04)
	AP propulsion GRF	0.24 (0.03)	0.25 (0.04)	0.20 (0.05)	0.24 (0.04)	0.25 (0.03)
Non-dominant Limb	**Vertical GRF early stance**	1.20 (0.08)	1.44 (0.14)	1.11 (0.09)	1.23 (0.11)	1.24 (0.11)
	Vertical GRF mid stance	0.70 (0.05)	0.66 (0.07)	0.70 (0.07)	0.65 (0.07)	0.68 (0.06)
	Vertical GRF late stance	1.12 (0.06)	0.87 (0.09)	1.16 (0.10)	1.10 (0.06)	1.12 (0.08)
	ML GRF early stance	-0.04 (0.02)	-0.03 (0.02)	-0.03 (0.02)	-0.03 (0.02)	-0.05 (0.03)
	ML GRF stance	0.08 (0.04)	0.11 (0.06)	0.06 (0.03)	0.08 (0.04)	0.09 (0.05)
	AP braking GRF	-0.21 (0.02)	-0.24 (0.05)	-0.18 (0.03)	-0.24 (0.03)	-0.22 (0.03)
	AP propulsion GRF	0.26 (0.03)	0.25 (0.04)	0.22 (0.05)	0.28 (0.04)	0.27 (0.04)

Table A17. Transtibial amputee peak ground reaction force, only for walking activities when the motion platform was stationary. Standard deviation is in brackets. Main effects for speed condition were observed for all parameters (boldfaced). However, significant differences between fixed and self-paced speed conditions were not significant with a treadmill speed covariate.

		Fixed Speed				
Transtibial Amputee		Level	Downhill	Uphill	Top-cross-slope	Bottom-cross-slope
Intact Limb	**Vertical GRF early stance**	1.15 (0.09)	1.48 (0.10)	1.23 (0.10)	1.18 (0.07)	1.19 (0.10)
	Vertical GRF mid stance	0.80 (0.06)	0.73 (0.07)	0.71 (0.09)	0.78 (0.08)	0.81 (0.08)
	Vertical GRF late stance	1.07 (0.07)	1.02 (0.08)	1.23 (0.09)	1.04 (0.06)	1.10 (0.05)
	ML GRF early stance	-0.06 (0.02)	-0.04 (0.02)	-0.04 (0.01)	-0.06 (0.02)	-0.06 (0.02)
	ML GRF stance	0.11 (0.03)	0.15 (0.05)	0.10 (0.03)	0.11 (0.02)	0.12 (0.03)
	AP braking GRF	-0.19 (0.04)	-0.26 (0.06)	-0.18 (0.03)	-0.20 (0.04)	-0.20 (0.03)
	AP propulsion GRF	0.20 (0.04)	0.21 (0.04)	0.25 (0.08)	0.21 (0.05)	0.22 (0.04)
Prosthetic Limb	**Vertical GRF early stance**	1.09 (0.11)	1.26 (0.19)	1.06 (0.12)	1.09 (0.09)	1.12 (0.11)
	Vertical GRF mid stance	0.82 (0.07)	0.82 (0.08)	0.77 (0.14)	0.81 (0.09)	0.83 (0.08)
	Vertical GRF late stance	1.04 (0.09)	0.94 (0.07)	1.03 (0.09)	1.03 (0.07)	1.03 (0.08)
	ML GRF early stance	-0.01 (0.01)	-0.02 (0.01)	-0.01 (0.01)	-0.01 (0.01)	-0.02 (0.01)
	ML GRF stance	0.09 (0.02)	0.10 (0.04)	0.09 (0.03)	0.09 (0.02)	0.09 (0.03)
	AP braking GRF	-0.13 (0.05)	-0.10 (0.06)	-0.17 (0.06)	-0.14 (0.04)	-0.12 (0.04)
	AP propulsion GRF	0.14 (0.03)	0.18 (0.04)	0.13 (0.04)	0.16 (0.03)	0.15 (0.03)
		Self-paced Speed				
Transtibial Amputee		Level	Downhill	Uphill	Top-cross-slope	Bottom-cross-slope
Intact Limb	**Vertical GRF early stance**	1.10 (0.36)	1.52 (0.14)	1.26 (0.09)	1.15 (0.37)	1.16 (0.38)
	Vertical GRF mid stance	0.70 (0.24)	0.68 (0.24)	0.68 (0.24)	0.68 (0.24)	0.68 (0.24)
	Vertical GRF late stance	1.12 (0.07)	1.06 (0.34)	1.06 (0.34)	1.06 (0.34)	1.06 (0.34)
	ML GRF early stance	-0.07 (0.02)	-0.06 (0.02)	-0.06 (0.02)	-0.06 (0.02)	-0.06 (0.02)
	ML GRF stance	0.12 (0.03)	0.12 (0.05)	0.12 (0.05)	0.12 (0.05)	0.12 (0.05)
	AP braking GRF	-0.23 (0.04)	-0.20 (0.08)	-0.20 (0.08)	-0.20 (0.08)	-0.20 (0.08)
	AP propulsion GRF	0.25 (0.06)	0.23 (0.09)	0.23 (0.09)	0.23 (0.09)	0.23 (0.09)
Prosthetic Limb	**Vertical GRF early stance**	1.14 (0.09)	1.29 (0.22)	1.05 (0.12)	1.06 (0.35)	1.21 (0.16)
	Vertical GRF mid stance	0.79 (0.08)	0.78 (0.09)	0.78 (0.09)	0.78 (0.09)	0.78 (0.09)
	Vertical GRF late stance	1.04 (0.09)	1.03 (0.10)	1.03 (0.10)	1.03 (0.10)	1.03 (0.10)
	ML GRF early stance	-0.01 (0.01)	-0.02 (0.01)	-0.02 (0.01)	-0.02 (0.01)	-0.02 (0.01)
	ML GRF stance	0.09 (0.02)	0.09 (0.02)	0.09 (0.02)	0.09 (0.02)	0.09 (0.02)
	AP braking GRF	-0.14 (0.05)	-0.16 (0.05)	-0.16 (0.05)	-0.16 (0.05)	-0.16 (0.05)
	AP propulsion GRF	0.16 (0.02)	0.17 (0.03)	0.17 (0.03)	0.17 (0.03)	0.17 (0.03)

Table A18. Able-bodied peak ground reaction force (GRF) mean of the standard deviations (mSD), only for walking activities when the motion platform was stationary. Standard deviation of the standard deviations is in brackets. Main effects for speed condition were observed for peak vertical and anterior-posterior GRF parameters (boldfaced). However, significant differences between fixed and self-paced speed conditions were not significant with a treadmill speed covariate.

				Fixed Speed		
Able-bodied		Level	Downhill	Uphill	Top-cross-slope	Bottom-cross-slope
Dominant Limb	**Vertical GRF early stance**	0.03 (0.01)	0.05 (0.02)	0.03 (0.01)	0.03 (0.02)	0.03 (0.01)
	Vertical GRF mid stance	0.03 (0.01)	0.03 (0.01)	0.03 (0.01)	0.02 (0.01)	0.02 (0.01)
	Vertical GRF late stance	0.02 (0.01)	0.04 (0.01)	0.03 (0.01)	0.02 (0.01)	0.03 (0.01)
	ML GRF early stance	0.01 (0.01)	0.01 (0.00)	0.01 (0.01)	0.01 (0.01)	0.01 (0.01)
	ML GRF stance	0.01 (0.01)	0.01 (0.01)	0.01 (0.01)	0.01 (0.01)	0.01 (0.01)
	AP braking GRF	0.02 (0.01)	0.03 (0.01)	0.02 (0.01)	0.02 (0.01)	0.02 (0.01)
	AP propulsion GRF	0.00 (0.00)	0.02 (0.00)	0.02 (0.01)	0.01 (0.01)	0.01 (0.01)
Non-dominant Limb	**Vertical GRF early stance**	0.03 (0.01)	0.05 (0.02)	0.04 (0.02)	0.04 (0.02)	0.04 (0.02)
	Vertical GRF mid stance	0.02 (0.01)	0.03 (0.01)	0.03 (0.01)	0.03 (0.02)	0.02 (0.01)
	Vertical GRF late stance	0.02 (0.01)	0.03 (0.01)	0.04 (0.02)	0.03 (0.03)	0.02 (0.02)
	ML GRF early stance	0.01 (0.01)	0.01 (0.01)	0.01 (0.01)	0.01 (0.01)	0.01 (0.01)
	ML GRF stance	0.01 (0.01)	0.01 (0.01)	0.01 (0.01)	0.01 (0.01)	0.01 (0.01)
	AP braking GRF	0.01 (0.01)	0.03 (0.01)	0.02 (0.01)	0.02 (0.01)	0.02 (0.01)
	AP propulsion GRF	0.00 (0.00)	0.02 (0.01)	0.02 (0.01)	0.01 (0.01)	0.01 (0.01)
				Self-paced Speed		
Able-bodied		Level	Downhill	Uphill	Top-cross-slope	Bottom-cross-slope
Dominant Limb	**Vertical GRF early stance**	0.05 (0.02)	0.06 (0.03)	0.04 (0.02)	0.04 (0.02)	0.04 (0.02)
	Vertical GRF mid stance	0.04 (0.02)	0.03 (0.02)	0.04 (0.01)	0.02 (0.01)	0.03 (0.02)
	Vertical GRF late stance	0.03 (0.01)	0.04 (0.02)	0.04 (0.02)	0.03 (0.02)	0.04 (0.02)
	ML GRF early stance	0.01 (0.01)	0.01 (0.01)	0.01 (0.01)	0.01 (0.01)	0.01 (0.01)
	ML GRF stance	0.01 (0.01)	0.02 (0.01)	0.01 (0.01)	0.01 (0.01)	0.01 (0.01)
	AP braking GRF	0.02 (0.01)	0.03 (0.01)	0.03 (0.01)	0.02 (0.01)	0.02 (0.01)
	AP propulsion GRF	0.02 (0.01)	0.02 (0.01)	0.02 (0.01)	0.02 (0.01)	0.02 (0.01)
Non-dominant Limb	**Vertical GRF early stance**	0.05 (0.02)	0.06 (0.04)	0.04 (0.02)	0.04 (0.03)	0.05 (0.04)
	Vertical GRF mid stance	0.04 (0.02)	0.04 (0.02)	0.04 (0.03)	0.03 (0.02)	0.02 (0.02)
	Vertical GRF late stance	0.03 (0.02)	0.04 (0.02)	0.04 (0.02)	0.03 (0.01)	0.03 (0.03)
	ML GRF early stance	0.01 (0.01)	0.01 (0.01)	0.01 (0.00)	0.01 (0.01)	0.01 (0.01)
	ML GRF stance	0.01 (0.01)	0.02 (0.01)	0.01 (0.01)	0.01 (0.01)	0.01 (0.01)
	AP braking GRF	0.02 (0.01)	0.02 (0.01)	0.02 (0.01)	0.02 (0.01)	0.02 (0.02)
	AP propulsion GRF	0.02 (0.01)	0.03 (0.01)	0.02 (0.01)	0.02 (0.01)	0.02 (0.01)

Table A19. Transtibial amputee peak ground reaction force (GRF) mean of the standard deviations (mSD), only for walking activities when the motion platform was stationary. Standard deviation of the standard deviations is in brackets. Main effects for speed condition were observed for peak vertical and anterior-posterior GRF parameters (boldfaced). When considering speed mSD as a covariate, only peak AP braking GRF was significantly smaller for self-paced compared to the fixed speed condition (only downhill for intact limb) and peak AP propulsion GRF was significantly larger for self-paced compared to the fixed speed condition (only bottom-cross-slope for intact limb).

				Fixed Speed		
Transtibial Amputee		Level	Downhill	Uphill	Top-cross-slope	Bottom-cross-slope
Intact Limb	**Vertical GRF early stance**	0.05 (0.03)	0.10 (0.04)	0.05 (0.02)	0.05 (0.02)	0.05 (0.02)
	Vertical GRF mid stance	0.04 (0.03)	0.04 (0.01)	0.04 (0.02)	0.03 (0.01)	0.04 (0.02)
	Vertical GRF late stance	0.04 (0.02)	0.04 (0.02)	0.04 (0.03)	0.03 (0.02)	0.03 (0.01)
	ML GRF early stance	0.01 (0.01)	0.02 (0.01)	0.01 (0.01)	0.01 (0.00)	0.01 (0.01)
	ML GRF stance	0.02 (0.01)	0.02 (0.01)	0.02 (0.01)	0.02 (0.01)	0.03 (0.01)
	AP braking GRF	0.02 (0.01)	**0.05 (0.02)**	0.03 (0.01)	0.03 (0.01)	0.03 (0.02)
	AP propulsion GRF	0.03 (0.01)	0.02 (0.01)	0.02 (0.02)	0.02 (0.01)	**0.01 (0.01)**
Prosthetic Limb	**Vertical GRF early stance**	0.04 (0.02)	0.07 (0.07)	0.04 (0.02)	0.04 (0.02)	0.05 (0.03)
	Vertical GRF mid stance	0.03 (0.01)	0.04 (0.02)	0.03 (0.02)	0.02 (0.01)	0.02 (0.01)
	Vertical GRF late stance	0.02 (0.01)	0.04 (0.02)	0.03 (0.02)	0.03 (0.01)	0.02 (0.01)
	ML GRF early stance	0.01 (0.01)	0.01 (0.00)	0.01 (0.01)	0.01 (0.01)	0.01 (0.00)
	ML GRF stance	0.01 (0.01)	0.02 (0.01)	0.02 (0.01)	0.01 (0.01)	0.02 (0.01)
	AP braking GRF	0.02 (0.01)	0.02 (0.01)	0.02 (0.01)	0.02 (0.01)	0.02 (0.01)
	AP propulsion GRF	0.02 (0.01)	0.02 (0.01)	0.02 (0.01)	0.02 (0.01)	0.01 (0.01)
				Self-paced Speed		
Transtibial Amputee		Level	Downhill	Uphill	Top-cross-slope	Bottom-cross-slope
Intact Limb	**Vertical GRF early stance**	0.04 (0.03)	0.09 (0.06)	0.07 (0.03)	0.06 (0.06)	0.05 (0.04)
	Vertical GRF mid stance	0.05 (0.02)	0.06 (0.04)	0.06 (0.04)	0.06 (0.04)	0.06 (0.04)
	Vertical GRF late stance	0.05 (0.04)	0.05 (0.03)	0.05 (0.03)	0.05 (0.03)	0.05 (0.03)
	ML GRF early stance	0.01 (0.01)	0.01 (0.01)	0.01 (0.01)	0.01 (0.01)	0.01 (0.01)
	ML GRF stance	0.02 (0.01)	0.02 (0.02)	0.02 (0.02)	0.02 (0.02)	0.02 (0.02)
	AP braking GRF	0.03 (0.02)	**0.04 (0.02)**	0.04 (0.02)	0.04 (0.02)	0.04 (0.02)
	AP propulsion GRF	0.02 (0.02)	0.03 (0.02)	0.03 (0.02)	0.03 (0.02)	**0.03 (0.02)**
Prosthetic Limb	**Vertical GRF early stance**	0.05 (0.03)	0.08 (0.06)	0.05 (0.03)	0.04 (0.03)	0.06 (0.04)
	Vertical GRF mid stance	0.04 (0.02)	0.04 (0.02)	0.04 (0.02)	0.04 (0.02)	0.04 (0.02)
	Vertical GRF late stance	0.03 (0.01)	0.03 (0.01)	0.03 (0.01)	0.03 (0.01)	0.03 (0.01)
	ML GRF early stance	0.01 (0.01)	0.01 (0.01)	0.01 (0.01)	0.01 (0.01)	0.01 (0.01)
	ML GRF stance	0.01 (0.01)	0.02 (0.01)	0.02 (0.01)	0.02 (0.01)	0.02 (0.01)
	AP braking GRF	0.02 (0.02)	0.02 (0.01)	0.02 (0.01)	0.02 (0.01)	0.02 (0.01)
	AP propulsion GRF	0.02 (0.01)	0.02 (0.01)	0.02 (0.01)	0.02 (0.01)	0.02 (0.01)

Appendix F: Amputee group information sheet

The Ottawa Hospital | L'Hôpital d'Ottawa

Information Sheet and Consent Form

Dynamic Stability Assessment within Rehabilitation Virtual Reality Environments for Improved Mobility

Principal Investigators: Dr. Markus Besemann, [redacted]
Dr. Edward Lemaire, [redacted]

Funding Source: Canadian Forces Health Service

<u>Introduction</u>

You are being asked to participate in this research project to measure dynamic stability performance while walking in a virtual reality (VR) system. The outcomes from this project will provide information that healthcare professionals can use to better understand walking stability, and thereby improve rehabilitation decision-making.

Please read this Information Sheet and Consent Form carefully and ask as many questions as you like before deciding whether to participate in this research study. You can discuss this decision with your family, friends and your health-care team.

<u>Background, Purpose and Design of the Study</u>

For people with amputations, understanding their dynamic stability is important when making healthcare decisions. Dynamic stability refers to stability during movement. In addition to the potential link between fall risk and critical instability (i.e., the point where the person becomes so unstable that they cannot maintain balance), dynamic stability could relate to movement confidence, enhanced performance, and enhanced quality of life.

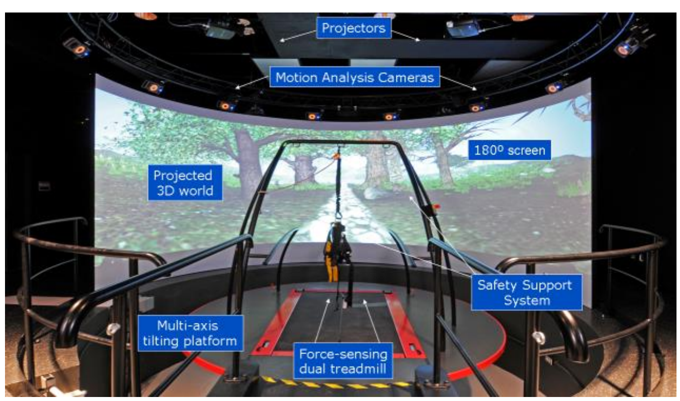

Figure 1: **CAREN Extended Virtual Reality System.**

A total of 25 individuals with lower extremity amputation and 15 individuals without lower extremity injury will be recruited for this study. This study will use new virtual reality technology (CAREN-Extended system, Figure 1), movement analysis, foot pressure analysis, and movement confidence and mobility questionnaires to develop dynamic stability measures for mobility assessment. The virtual reality system will provide 3D environments so that a study participant's dynamic stability is challenged as they walk through the computer-generated world.

You are eligible to participate in this project if you have one transtibial or transfemoral amputation. You must have completed physical rehabilitation, use your prosthesis daily, and have successfully completed a gait training program. Full-body movements and foot pressures/forces will be recorded while you walk on a treadmill within the CAREN-System. The scenes include a path that moves up and down, side to side, and multiple uneven ground conditions (similar to walking down an uneven sidewalk, cobblestone, or small hill, etc.). The information collected while you walk through each 3D scene will be analyzed to help us understand which measures and scene designs are useful for understanding dynamic stability.

Study Procedures and or Description of Treatment

All data collection will take place within the Rehabilitation Virtual Reality (RVR) lab at The Ottawa Hospital Rehabilitation Centre (TOHRC). Upon arriving at the RVR lab, a project assistant will review the protocol and ask you to complete the consent form. A prosthetist will check your prosthesis and residual limb to ensure that the device is functioning appropriately and that your residual limb is healthy. You will then be asked to complete the Activities-specific Balance Confidence scale, a questionnaire which asks you to rate confidence and ease for common activities of daily living, and the Lower Extremity Functional Status Survey, which asks you to rate how easy it is to perform various activities. These questionnaires should take approximately 15 minutes to complete. A project assistant will record your age, height, weight, sex, amputation level, cause of amputation, and prosthetic components.

You will be asked to wear a tight fitting short sleeve shirt or tank top, athletic shorts, and athletic shoes to minimize marker movement during testing. A project assistant will attach reflective markers to your feet, legs, pelvis, trunk, arms, and head so that we can track how you move. A pressure sensor insole will be fitted to your shoe and attached to a belt pack, worn at your waist, to measure in-shoe foot pressure as you walk.

The CAREN-Extended system has a safety support attachment so that you do not fall to the ground during a stumble. A safety harness will be adjusted to you and attached to the support structure before the system is started. The CAREN system operator will explain all safety procedures during the session.

To measure your dynamic stability performance, you will walk through a virtual park scene with slopes, side-to-side motion, and uneven ground walking. To become accustomed to the CAREN system, you will practice each walking condition present in the virtual reality scene. The practice trials will take approximately 15 minutes. After you've completed the practice trials, you will be asked to repeat this application three times at two walking speeds for a total of six trials. Each trial will take approximately 6 minutes to complete. You may rest at any time between trials. You can also have the system stopped at any time during a trial.

Study Duration

The entire test session is expected to last 2.5 hours with approximately 30-40 minutes of walking. After this session, your participation in this study will be complete.

Possible Side Effects and/or Risks

Since you will be wearing the safety harness, risks in the study are minimal and may include chaffing from the harness or discomfort if you fall and the harness stops you abruptly.

Some people experience dizziness when working with a VR system. If this occurs during the simple VR applications, more time can be spent allowing you to become accustomed to the system or you can choose to withdraw from the study. If you feel uncomfortable during any activity, the system can be stopped at your request.

Benefits of the Study

You may not receive any direct benefit from your participating in this study. Your participation in this research may allow the researchers to discover better ways to use VR systems to improve our understanding of dynamic stability, leading to better decisions for prosthetic prescription, prosthesis setup, or therapy related amputee movement.

Withdrawal from the Study

You have the right to withdraw from the study at any time without any impact to your current and future care at the Ottawa Hospital. If you decide to withdraw, you should discuss this with the lead investigator and provide an explanation to the research team.

You have the right to check your study records and request changes if the information is not correct. However, to ensure the scientific integrity of the study, some of your records related to the study may not be available for your review until after the study has been completed.

Compensation

In the event of a research-related injury or illness, you will be provided with appropriate medical treatment/care. You are not waiving your legal rights by agreeing to participate in this study. The study doctor and the hospital still have their legal and professional responsibilities.

Study Costs

Your parking costs at TOHRC will be paid while participating in the study.

Confidentiality

All personal health information will be kept confidential, unless release is required by law. Representatives of the Ottawa Hospital Research Ethics Board, as well as the Ottawa Hospital Research Institute, may review your original research records under the supervision of Dr. Edward Lemaire's staff for audit purposes.

You will not be identifiable in any publications or presentations resulting from this study. No identifying information will leave the Ottawa Hospital. All information which leaves the hospital will be coded with an independent study number.

The link between your name and the independent study number will only be accessible by Dr. Edward Lemaire and/or his staff. The link and study files will be stored separately and securely. Both files will be kept for a period of 10 years after the study has been completed. All paper records will be stored in a locked file and/or office. All electronic records will be stored on The Ottawa Hospital network and protected by a user password, again only accessible by Dr. Edward Lemaire and/or his staff. At the end of the retention period, all paper records will be disposed of in confidential waste or shredded, and all electronic records will be deleted.

Voluntary Participation

Your participation in this study is voluntary. If you choose not to participate, your decision will not affect the care you receive at this Institution at this time, or in the future. You will not have any penalty or loss of benefits to which you are otherwise entitled to.

New Information About the Study

You will be told of any new findings during the study that may affect your willingness to continue to participate in this study. You may be asked to sign a new consent form.

Questions about the Study

If you have any questions about this study or if you feel that you have experienced a research-related injury, please contact Dr. Edward Lemaire, [redacted].

The Ottawa Hospital Research Ethics Board (OHREB) has reviewed this protocol. The OHREB considers the ethical aspects of all research studies involving human subjects at The Ottawa Hospital. If you have any questions about your rights as a research subject, you may contact the Chairperson of the Ottawa Hospital Research Ethics Board at [redacted].

Appendix G: Control group information sheet

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Control Group Information Sheet and Consent Form

Dynamic Stability Assessment within Rehabilitation Virtual Reality Environments for Improved Mobility

Principal Investigators:	Dr. Markus Besemann, [redacted] Dr. Edward Lemaire, [redacted]
Funding Source:	Canadian Forces Health Service

<u>Introduction</u>

You are being asked to participate in this research project as a comparison control group to individuals with lower extremity amputation. This research project will measure dynamic stability performance during walking using a new virtual reality (VR) system. The outcomes from this project will provide information that healthcare professionals can use to better understand walking stability, and thereby improve rehab decision-making.

Please read this Information Sheet and Consent Form carefully and ask as many questions as you like before deciding whether to participate in this research study. You can discuss this decision with your family, friends and your health-care team.

<u>Background, Purpose and Design of the Study</u>

For people with amputations, understanding their dynamic stability is important when making healthcare decisions. Dynamic stability refers to stability during movement. In addition to the potential link between fall risk and critical instability (i.e., the point where the person becomes so unstable that they cannot maintain balance), dynamic stability could relate to movement confidence, enhanced performance, and enhanced quality of life.

A total of 25 individuals with lower extremity amputation and 15 individuals without lower extremity injury will be recruited for this study. This study will use new virtual reality technology (CAREN-Extended system, Figure 1), movement analysis, foot pressure analysis, and movement confidence and mobility questionnaires to develop dynamic stability measures for mobility assessment. The virtual reality system will provide 3D environments so that a study participant's dynamic stability is challenged as they walk through the computer-generated world.

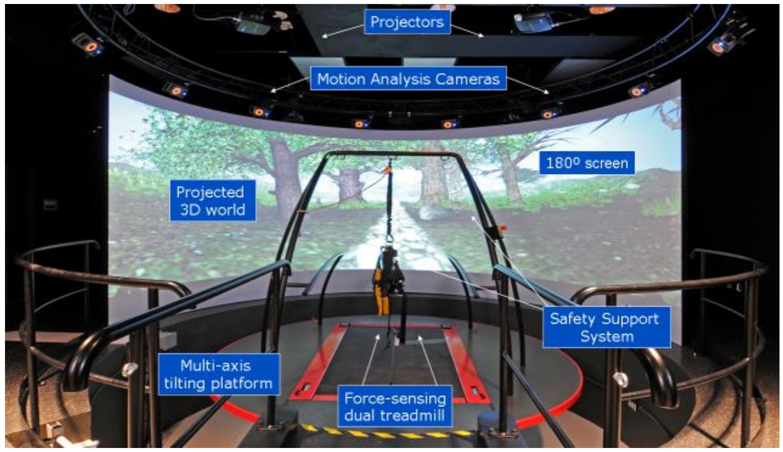

Figure 1: **CAREN Extended Virtual Reality System.**

You are eligible to participate in this study if you have not sustained an injury (or surgery) to a lower extremity, and if you are a similar age and sex to the amputee group. Full-body movements and foot pressures/forces will be recorded while a person walks on a treadmill within the CAREN-System. The scenes include a path that moves up and down, side to side, and multiple uneven ground conditions (similar to walking down an uneven sidewalk, cobblestone, or small hill, etc.). The information collected while you walk through each 3D scene will be analyzed to help us understand which measures and scene designs are useful for understanding dynamic stability.

Study Procedures and or Description of Treatment

All data collection will take place within the Rehabilitation Virtual Reality (RVR) lab at The Ottawa Hospital Rehabilitation Centre (TOHRC). Upon arriving at the RVR lab, a project assistant will review the protocol and ask you to complete the consent form. A project assistant will record your age, height, weight, and sex.

You will be asked to wear a tight fitting short sleeve shirt or tank top, athletic shorts, and athletic shoes to minimize marker movement during testing. A project assistant will attach reflective markers to your feet, legs, pelvis, trunk, arms, and head so that we can track how you move. A pressure sensor insole will be fitted to your shoe and attached to a belt pack, worn at your waist, to measure in-shoe foot pressure as you walk.

The CAREN-Extended system has a safety support attachment so that you do not fall to the ground during a stumble. A safety harness will be adjusted to you and attached to the support structure before the system is started. The CAREN system operator will explain all safety procedures during the session.

To measure your dynamic stability performance, you will walk through a virtual park scene with slopes, side-to-side motion, and uneven ground walking. To become accustomed to the CAREN system, you will practice each walking condition present in the virtual reality scene. The practice trials will take approximately 15 minutes. After you've completed the practice trials, you will be asked to repeat this application three times at two walking speeds for a total of six trials. Each trial will take approximately 6 minutes to complete. You may rest at any time between trials. You can also have the system stopped at any time during a trial.

Study Duration

The entire test session is expected to last 2.5 hours with approximately 30-40 minutes of walking. After this session, your participation in this study will be complete.

Possible Side Effects and/or Risks

Since you will be wearing the safety harness, risks in the study are minimal and may include chaffing from the harness or discomfort if you fall and the harness stops you abruptly.

Some people experience dizziness when working with a VR system. If this occurs during the simple VR applications, more time can be spent allowing you to become accustomed to the system or you can choose to withdraw from the study. If you feel uncomfortable during any activity, the system can be stopped at your request.

Benefits of the Study

You may not receive any direct benefit from your participating in this study. Your participation in this research may allow the researchers to discover better ways to use VR systems to improve our understanding of dynamic stability, leading to better decisions for prosthetic prescription, prosthesis setup, or therapy related amputee movement.

Withdrawal from the Study

You have the right to withdraw from the study at any time without any impact to your current and future care at the Ottawa Hospital. If you decide to withdraw, you should discuss this with the lead investigator and provide an explanation to the research team.

You have the right to check your study records and request changes if the information is not correct. However, to ensure the scientific integrity of the study, some of your records related to the study may not be available for your review until after the study has been completed.

Compensation

In the event of a research-related injury or illness, you will be provided with appropriate medical treatment/care. You are not waiving your legal rights by agreeing to participate in this study. The study doctor and the hospital still have their legal and professional responsibilities.

Study Costs

Your parking costs at TOHRC will be paid while participating in the study.

Confidentiality

All personal health information will be kept confidential, unless release is required by law. Representatives of the Ottawa Hospital Research Ethics Board, as well as the Ottawa Hospital Research Institute, may review your original research records under the supervision of Dr. Edward Lemaire's staff for audit purposes.

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Voluntary Participation

Your participation in this study is voluntary. If you choose not to participate, your decision will not affect the care you receive at this Institution at this time, or in the future. You will not have any penalty or loss of benefits to which you are otherwise entitled to.

New Information About the Study

You will be told of any new findings during the study that may affect your willingness to continue to participate in this study. You may be asked to sign a new consent form.

Questions about the Study

If you have any questions about this study or if you feel that you have experienced a research-related injury, please contact Dr. Edward Lemaire, [redacted].

The Ottawa Hospital Research Ethics Board (OHREB) has reviewed this protocol. The OHREB considers the ethical aspects of all research studies involving human subjects at The Ottawa Hospital. If you have any questions about your rights as a research subject, you may contact the Chairperson of the Ottawa Hospital Research Ethics Board at [redacted].

Appendix H: Consent form

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Consent Form

Dynamic Stability Assessment within Rehabilitation Virtual Reality Environments for Improved Mobility

Consent to Participate in Research

I understand that I am being asked to participate in a research study about using a new virtual reality (VR) application to measure stability while people move. This study has been explained to me by a member of the project team.

I have read this five-page Patient Information Sheet and Consent Form (or have had this document read to me). All my questions have been answered to my satisfaction. If I decide at a later stage in the study that I would like to withdraw my consent, I may do so at any time.

I voluntarily agree to participate in this study.

A copy of the signed Information Sheet and/or Consent Form will be provided to me.

Signatures

____
Participant's Name (Please Print)

______________________________ _________________
Participant's Signature Date

Investigator Statement (or Person Explaining the Consent)

I have carefully explained to the research participant the nature of the above research study. To the best of my knowledge, the research participant signing this consent form understands the nature, demands, risks and benefits involved in participating in this study. I acknowledge my responsibility for the care and well being of the above research participant, to respect the rights and wishes of the research participant, and to conduct the study according to applicable Good Clinical Practice guidelines and regulations.

____
Name of Investigator/Delegate (Please Print)

____________________________________ _________________
Signature of Investigator/Delegate Date