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Full Length Article

Benchmarking System for Assessing Balance (BeStABLE) during gait using visual stepping cues

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ABSTRACT

Background: The ability to characterise balance capacity through clinical assessment procedures is often limited by a focus on movement tasks substantially more predictable than those experienced in the real world. Consequently, assessment protocols may not sufficiently elicit and characterise the full range of sensory, cognitive and physical challenges present in daily life.

Methods: A task-specific procedure for assessing balance control during gait via a treadmill-based augmented-reality environment called the 'Benchmarking System for Assessing Balance' (BeStABLE) is proposed. The apparatus utilises unpredictable visual cues that prescribe discrete changes to foot placement patterns, requiring rapid stimulus perception, decision-making, and modification of body dynamics, followed by compensatory corrections to return to steady gait. This is representative of a wide variety of situations in community living and many reported circumstances of real-world falls. This assessment procedure is demonstrated with a case study comparing three high-functioning users of transtibial (below-knee) prostheses with a group of 19 able-bodied persons.

Results: The BeStABLE successfully discriminates the prosthesis users from the normative group in multiple outcomes, and most prominently when participants are cued to narrow their steps. The finding that certain gait adaptations are not realisable by the prosthesis users may help to refine future protocols for assessing gait and balance impairment with the BeStABLE.

Conclusion: The BeStABLE treadmill-based augmented-reality environment enables the repeatable and objective assessment of gait adaptations through a process that is safe for subjects and efficient for clinicians.

1. Introduction

Balance control is the proactive and reactive regulation of the motion of the elevated centre of mass to avoid falling and entails a flexible interaction of multiple sensory, cognitive, and motor sub-processes that change with the situation [1,2], such as stepping over or around obstacles or recovering equilibrium after tripping. This contextual dependency poses a considerable challenge for characterising balance function because norm-referenced performance in one situation can vary substantially from another. The combined cognitive and physical challenges of navigating unpredictable environments partly explain why walking is the activity most frequently associated with falling amongst high-risk populations [3].

Individuals with a unilateral transtibial amputation utilise a number of compensation strategies to realise gait. The lack in ankle function

limits the ability to shift the centre of pressure in the anterior direction to quickly modulate propulsive and braking forces, and so gait adaptations during the stance phase of the non-affected leg are used [4,5]. More broadly, the kinematic and dynamic characteristics of the left and right foot contact events become increasingly asymmetric in the stride length, stance duration, and ground-reaction forces [6], and as most of the load is shifted onto the non-affected leg, amputees exhibit higher mediolateral sway [7].

Though several existing protocols use a variety of testing conditions to isolate and characterise multiple relevant subsystems, there remains no gold standard for comprehensively quantifying balance performance [8]. It can be argued that standard clinical assessments of balance and mobility are often too structured to extensively probe

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reactive adaptations and rarely include tasks with unpredictable stimuli that necessitate non-stereotyped, context-dependent responses. Instead, they usually emphasise repeatability, for example, standing in static postures – which clearly limit the capture of dynamic balance control and reactive adaptations – or executing pre-determined transitions between sitting, standing and walking.

In addition to inconsistencies between different clinical scales [9], research has shown that approaches such as the Berg Balance Scale (BBS) and the Timed Up and Go (TUG) test yield inconsistent results, while other approaches emphasising gait speed are often poor predictors of fall likelihood [10]. Hence, current clinical balance assessment procedures often fail to sufficiently predict fall risk or discriminate fallers from non-fallers [11], and less than 30% of clinicians report that existing balance assessment tools are adequate for their needs [12]. However, effective assessment is critical for detecting balance disorders, guiding interventions, and evaluating progress, while unreliable assessment may mischaracterise individuals with balance impairments, lead to the premature release of patients from care providers and thereby indirectly contribute to the incidence of falls and consequent anxiety, injury, or death [13].

In recent years, visually-cued stepping and walking protocols have emerged as a task-specific and economical means of assessing goal-directed gait adaptability in laboratory studies. External cues are frequently used in rehabilitation and can have visual, auditory or tactile forms. Visual cues are often applied as markers or laser pointers at various distances from the subject [14]. These protocols emulate situations of balance recovery and obstacle avoidance by suddenly presenting stepping obstacles or targets on the walking surface, often represented by embedded lights [15], screens [16], or optical projections [17]. Since amputees tend to be more dependent on visual feedback for both quiet standing and gait, sway and quadriceps activity during gait can be reduced through visual feedback [18] thanks to the additional spatial information provided [19]. In addition to being applied for gait and balance assessment, visual cues have been used in gait training aimed at improving walking performance, for example, for people with Parkinson's Disease [20].

In addition to visually-cued stepping methods, augmented reality (AR) systems use computer-generated objects to create an interactive experience, in which users can adjust their actions according to real-time feedback. It has been argued that the platforms provide a more motivating and engaging environment and that they are well-suited to repetitive practise of key aspects of gait and balance [21]. AR platforms can also provide different trajectories to follow and obstacle avoidance, which can aid in balance training and create real-world gait training exercises and scenarios [22]. The technology can also facilitate greater task variability, which has been demonstrated to be important for motor learning and rehabilitation [23].

To complement existing balance assessment procedures which utilise the visually-cued and augmented reality techniques described above, we propose here a task-specific procedure for assessing the proactive and reactive control of balance by means of visual cues that prescribe unpredictable intermittent gait adaptations. An assessment platform called the Benchmarking System for Assessing Balance (BeStABLE) – which comprises a sensorised treadmill and augmented reality environment – is used in this research. We study the spatiotemporal gait characteristics of individuals both with and without transtibial amputation. The main advantages of the system, as well as arguably providing the subjects with more realistic and challenging gait tasks, are its low effort for setup and short measurement times. The main objectives of the study are to investigate whether various stepping targets at diverse locations for able-bodied (AB) subjects can lead to different stepping 'success' rates as a means of assessing gait performance. Furthermore, differences in the stepping characteristics between the AB subjects and three transtibial amputees are examined as a proof-of-concept study into the utility of the BeStABLE system as a means of assessing gait and balance in this group.

Table 1

Description of cues provided to subjects.

Cue	Description
W	wide
N	narrow
L	long
LW	long-wide
LN	long-narrow

2. Methods

2.1. Technical realisation

The Benchmarking System for Assessing Balance (BeStABLE, Fig. 1) is a treadmill-based augmented-reality environment that displays single stepping cues (targets) on the ground intermittently during otherwise uncued (unprescribed) steady gait at fixed belt speeds. The cues are quasi-randomly selected from a closed set of timings (an integer number of strides between cues) and foot placement deviations from nominal gait: wide (W), narrow (N), long (L), long-wide (LW), and long-narrow (LN) steps. The foot placement deviations, which constitute self-induced balance perturbations, are summarised in Table 1.

The BeStABLE architecture was selected because context-specific proactive adaptations require situational awareness, derived from mixed sensory stimuli but often predominantly vision. Moreover, it builds upon familiar multi-purpose equipment commonly used in physiotherapy.

The platform comprises a single-belt instrumented treadmill (N-Mill Research, Motek Medical BV, Amsterdam, NL), ground-facing optical projection system (XJ-UT310WN, Casio Co. Ltd., Tokyo, Japan), and command PC running real-time instrument control and animation software (D-Flow, Motek Medical BV, Amsterdam, NL). Vertical force measurements from six transducers within the treadmill are communicated via a data acquisition board (USB-6210, National Instruments Corp., Austin, USA) to the command PC and real-time software to compute the total CoP of the test subject at a rate of 300 Hz.

For safety purposes, the Motek N-Mill has handrails on each side of the treadmill and two safety switches that immediately stop the treadmill movement: one is attached to a treadmill handrail for the participant and another to the PC console for the therapist. A ceiling-mounted safety harness prevents the subject from falling but otherwise provides no support.

The cues displayed by the BeStABLE are synchronised with the gait of the test subject. After optional randomisation with respect to the stride within which the cue appears, the stepping limb (left, right), and the type of cue (W, N, L, LW, or LN), the cue is timed to appear at the start of the swing phase of the selected stepping limb to yield a consistent response time. The selected cue is then placed relative to the predicted normal stepping location of the subject, and offset by a distance proportional to their leg length to present an equivalent challenge for individuals of different sizes (Fig. 1(b)).

Cues remain fixed with respect to the inertial frame and do not move with the treadmill surface. For task clarity, the actual target colours differ only for left (blue) and right (red) stepping limbs, and the desired stepping limb was further coded via the synchronised playing of a voice recording saying 'left' or 'right'. These synchronisations are achieved through the real-time estimation of gait events from the whole-body CoP trajectory recorded by the treadmill (Fig. 2). The gait events are detected as coincident with local extrema in the anteroposterior and mediolateral components of the CoP trajectory, as in previous studies by the authors (e.g. [24–30]) and others [31].

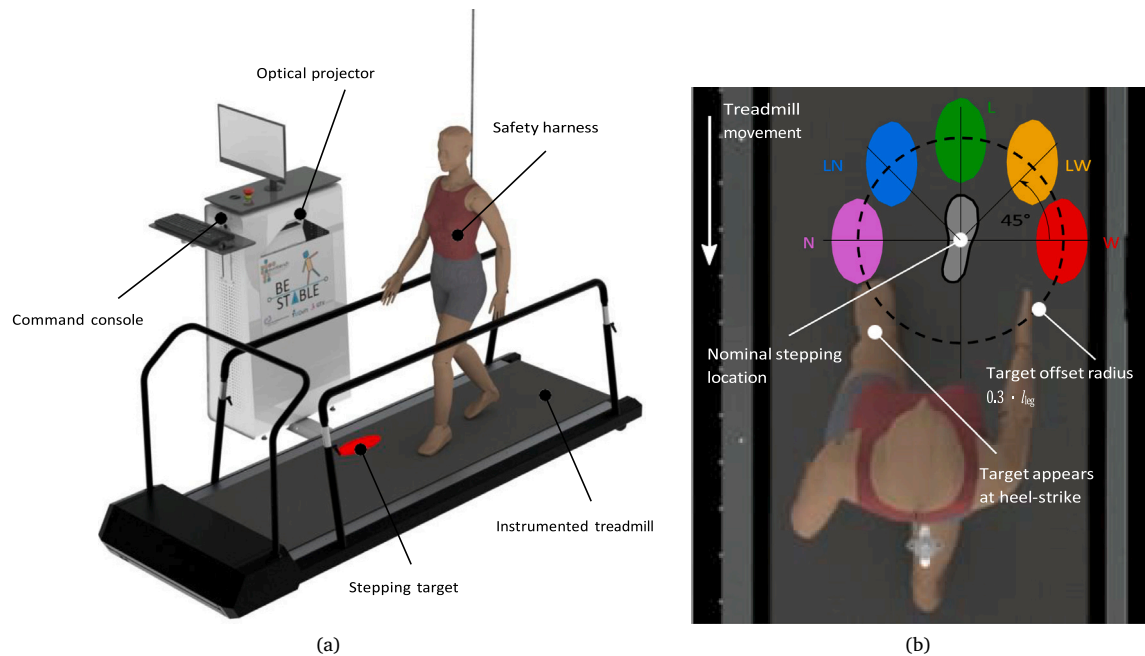


Fig. 1. (a) Components and application of the BeStABLE. (b) At the onset of the preceding stance phase of the stepping leg, an elliptical target appears on the treadmill surface in one of five randomly selected locations, cuing a W, N, L, LW, or LN stepping adaptation. The target position is relative to the prior uncued stepping locations and the subject's leg length (l_{leg}).

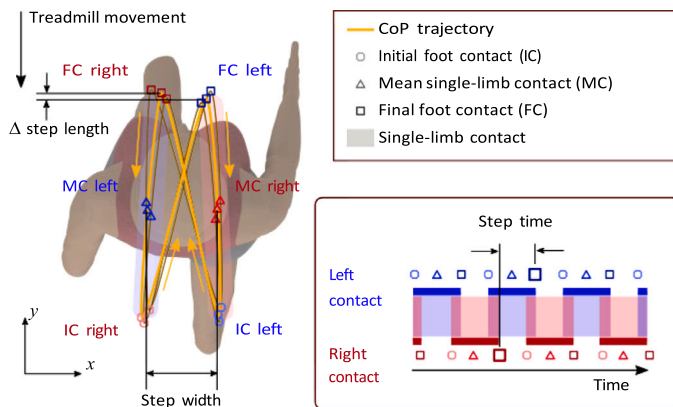


Fig. 2. Typical treadmill CoP trajectory showing the definitions of gait events and derived gait parameters.

2.2. Participants

Three adults with a unilateral transtibial amputation participated in the study; their characteristics are presented in Table 2. All the subjects were male and aged 48(9) years,¹ height 180(4) cm, mass 86(4) kg, all left-sided amputation, time since amputation 20(2.5) years and 19 non-amputee control subjects (8 females, age 38(11) years, height 176(7) cm, mass 74(13) kg) without any known musculoskeletal, orthopaedic or other neurological injuries. All amputee participants (A) were proficient prosthesis users and walked on their own without further walking aids or assistive devices. Unlike subjects A1 and A2, subject A3 had his prosthesis attached over the knee, locking the knee joint.

¹ $\bar{x}(\sigma_x)$, stating the mean ($\bar{x}$) and population standard deviation (σ_x) of a distributed quantity x .

The measurements were performed at the Research and Development department of the University Rehabilitation Institute of the Republic of Slovenia. The study was approved by the National Medical Ethics Committee Republic of Slovenia and all participants were briefed on the experimental protocol and signed an informed consent statement before being taking part.

2.3. Experimental protocol

The operator of the experiment entered the height of the participant in the BeStABLE console to personalise the magnitude of the prescribed gait adaptations. Next, the participants were instructed to step on a central area of the treadmill, where they were attached to the safety harness to prevent them from falling in case they lost their balance. They were further instructed to walk within a visually marked central area of the treadmill during the experiment and, when a visual cue was projected onto the treadmill surface, to try to step onto it with the foot currently in the swing phase.

They were then given a 5–10 min familiarisation session prior to data recording to acclimate to the treadmill speed in order to limit startle responses and to become acquainted with the cues. A 2 min session of uncued walking at a speed of 0.4 m/s was then recorded to establish the baseline gait characteristics.

The experimental protocol consisted in sequential blocks of three different treadmill speeds (0.4 m/s, 0.6 m/s, 0.8 m/s), each comprising a sequence of 50 randomly-ordered and randomly-occurring stepping targets (five directions $\times$ two limbs $\times$ 5 repetitions) and lasting about 10 min.

Each cue was followed by a wash-out period of randomly chosen 6–12 steps before the next cue was presented. After completion of the first block, a short break was taken to avoid fatigue since the participants were constantly observing the walking floor projections. After the break, the protocol was repeated for the walking speeds of 0.6 m/s and 0.8 m/s without familiarisation sessions for the new treadmill speeds.

The whole healthy control group was able to perform the protocol at all three walking speeds and for all five target locations. However, none of the amputee participants were presented with the ‘narrower’ target

Table 2
Participant data of transtibial amputee cohort.

Subject	Age (years)	Height (m)	Mass (kg)	Gender (M/F)	Amputation side (L/R)	Time since amputation (years)	Prosthesis ankle type ^a	Prosthesis fixation ^b
A1	48	1.78	88	M	L	22	uni-axis	below-knee
A2	60	1.86	80	M	L	20	uni-axis	below-knee
A3	37	1.76	90	M	L	25	multi-axis	above-knee

^a All prostheses had an articulated ankle designed to store and release energy. Uni-axis models support plantarflexion/dorsiflexion and multi-axis models additionally support eversion/inversion.

^b Prosthesis fixation below or above the intact knee joint. When attached above the knee, the prosthesis constrained the knee joint in a locked configuration.

location as it proved to be too challenging during the familiarisation trials. Furthermore, A1 and A2 performed the protocol at only the lowest walking speed (0.4 m/s).

2.4. Performance measures

The primary measure of gait adaptability is the proportion of each type of prescribed adaptation that is successfully performed, interpreted in relation to a normative data set of unimpaired individuals. Due to limb length differences, it will generally be easier for taller subjects to reach a target at a given distance away. Accordingly, in this research, the target radius used to assess stepping success is scaled by a factor corresponding to the subject's height normalised by the average height of the subjects. Hence, for a participant i of height h_i and an average cohort height $\bar{h} = 1.76$ m, an adaptation is defined to be successful if the stepping foot is placed within a radius r of the centre of the presented cue, where

$$r = \left(\frac{h_i}{\bar{h}} \right) \cdot 0.2 \quad (1)$$

and the foot placement is measured as the nearest CoP across the subsequent single-support phase. Secondary performance measures comprise spatio-temporal characteristics of the steps before, during, and after these adaptations. The quality of the adaptation is measured as the Euclidean error between the foot position (single-support CoP) and the centre of the cue.

The 0.2 m value was selected empirically based on preliminary testing. It was found to be sufficiently large to accommodate natural variability in foot placement, yet sufficiently small to ensure meaningful task accuracy and to avoid overlap with the nominal (uncued) stepping location.

Open-loop proactive adaptations are inferred from changes in the gait parameters between before and after exposure to cues, including the magnitudes of the durations, lengths, and widths of steps, as well as the variabilities and degrees of asymmetry in these magnitudes. Herein, the step time is the duration between the starts of two consecutive single-support phases.

Similarly, the step length is the difference in anteroposterior CoP positions (expressed in a belt-fixed coordinate frame) at these same instances. The step width is defined as the difference between the mean CoP positions across two consecutive single-support phases in order to account for the non-monotonicity of the lateral CoP displacement.

Following a cued step, reactive adaptations during the return to steady gait are assessed by comparing the step time, length, and width magnitudes of the three steps following the cued adaptation with the step preceding the cue.

2.5. Centre of pressure calculation

All performance measures are derived from the whole-body CoP trajectory recorded by the treadmill. Temporal gait events are inferred from the CoP trajectory with the same methodology used to synchronise the cues with the gait of the test subject. The CoP was obtained from the treadmill's embedded force-measurement system and real-time software environment [32]. The instrumented treadmill internally

combines the signals from six vertical-force transducers, whose spatial positions within the treadmill structure are known, to determine the resultant vertical ground reaction force and the corresponding CoP. The underlying principle is the standard weighted-average formulation for CoP estimation from multiple single-axis load cells. If the six transducers measure vertical forces $F_{z,j}$ at known positions (x_j, y_j) , then the CoP coordinates are obtained as

$$x_{CoP} = \frac{\sum_{i=1}^6 x_i F_{z,i}}{\sum_{i=1}^6 F_{z,i}} \quad (2)$$

and

$$y_{CoP} = \frac{\sum_{i=1}^6 y_i F_{z,i}}{\sum_{i=1}^6 F_{z,i}} \quad (3)$$

Foot position is quantified only during the single-support phase of the gait cycle and equivalent to the location of the CoP. In contrast to other definitions of foot position using kinematic landmarks (as in optical motion capture), the CoP varies within the base of support of the foot as ankle moments are exerted to manipulate the ground reaction forces and achieve stable locomotion [33,34].

2.6. Statistical analysis

Prior to analysis, spatial outcome measures were non-dimensionalised through division of the leg length of the subject ($l_{0,i} \approx 0.53h_i$, where h_i is the total body height of subject i) and temporal measures by multiplication by the natural frequency of the CoM modelled as a point-mass rotating as a linearised inverted pendulum about the ground [35,36] i.e.

$$\omega_{0,i} \approx \left(\frac{g}{0.528h_i} \right)^{\frac{1}{2}} \quad (4)$$

Further processing did not include explicit outlier detection or an exclusion scheme. Instead, repeated measurements with matching experimental conditions were aggregated per subject through computation of the median.

Multi-factor parametric repeated-measures analysis of variance (rANOVA) tests were applied to detect the presence of global trends as a prerequisite to investigating specific pairwise differences with appropriate parametric (t-test) or non-parametric (Wilcoxon signed-rank) repeated-measures tests. Statistical assumptions regarding normality and sphericity were evaluated using the Shapiro-Wilk and Mauchly tests, respectively. The latter were adjusted with the Greenhouse-Geisser and/or Huynh-Feldt methods depending on the magnitude of the violation [37]. To control for the inflation of Type-I errors during repeated posthoc comparisons, Bonferroni adjustments were made to the p-values and evaluated with a fixed significance threshold at $p = 0.05$.

However, these inferential statistical procedures were applied only to the healthy control group that completed all experimental conditions. The prosthesis users ($n = 3$) did not complete all conditions (e.g. narrower targets and higher walking speeds), and therefore did not provide a complete dataset suitable for repeated-measures statistical analysis. Moreover, the small sample size does not support reliable

verification of parametric assumptions. Consequently, no inferential statistical analyses were performed on the amputee dataset. Instead, the data from the prosthesis users were analysed descriptively; the intended role of the amputee cohort is to enable a case study illustrating the discriminatory capability of the methodology rather than supporting population-level inference.

Inferential statistical analyses were performed on the healthy control group ($n = 19$), which completed all experimental conditions (walking speeds and cue directions). Experimental (repeated-measures) factors relevant for analysis included treadmill speed ($speed \in \{0.4, 0.6, 0.8\} \text{ m/s}$), measurement session ($session \in \{\text{uncued, intermittently cued}\}$), stepping limb ($limb \in \{\text{left, right}\}$), direction of the cued adaptation ($direction \in \{W, LW, L, LN, N\}$), and step relative to a cue ($step \in \{\text{uncued (U), target cue (T), subsequent three recovery steps (R}_1\text{--R}_3\text{)}\}$). Due to a lack of significant differences in $limb$ during uncued walking, the left and right measurements were aggregated using the mean for all further analysis. Cued adaptation performance was assessed via two-way ($speed \times direction$) rANOVAs for the target contact success rate and magnitude of the target error. In the latter case, participants were excluded if they were unable to successfully contact every target at least once at each walking speed, which was the case for C14 and C17. Post-hoc analysis was performed across direction with speed aggregated, and across speed grouped by direction ($n = 25$ comparisons).

Gait adaptations made in anticipation of stepping cues were assessed via a two-way ($speed \times session$) rANOVA on step width, step length, step time, and their variability and symmetry indices, followed by orthogonal pairwise comparisons across session, grouped by speed ($n = 27$ comparisons). Individual steps during cued adaptations were assessed with a four-way ($speed \times limb \times direction \times step$) rANOVA on step width, step length, and step time, with specific differences compared across step grouped by $speed \times direction$ ($n = 180$ comparisons). The amputee group ($n = 3$) did not complete all experimental conditions (e.g. narrower cues and higher walking speeds were not feasible for some participants). Again, due to the small sample size and incomplete dataset, the assumptions required for parametric statistical testing were not satisfied; therefore, no inferential statistical analyses were performed on this cohort.

Cued adaptation performance was assessed via two-way ($speed \times direction$) rANOVAs for the target contact success rate and magnitude of the target error. In the latter case, participants were excluded if they were unable to successfully contact every target at least once at each walking speed, which was the case for C14 and C17. Post-hoc analysis was performed across direction with speed aggregated, and across speed grouped by direction ($n = 25$ comparisons).

As a diagnostic tool for detecting impairments in individuals, the BeStable compares limited measurements of a subject with a normative distribution approximating a general population of unimpaired individuals. The cumulative distribution function estimates the probability that a given measurement comes from this distribution and therefore represents an unimpaired individual. For this case study, the 50%, 75%, and 95% central quantile ranges of the normative population were used as benchmarks.

3. Results

3.1. Adaptation performance

Fig. 3 shows aggregated left and right data of the control group corresponding to the success rate and stepping error for the prescribed cue locations and walking speeds. For both the target-stepping success rate and target error, a significant effect was found for the able-bodied subjects for target location ($F(4, 72)=12.1$, $p<0.001$; $F(4, 64)=14.6$, $p<0.001$, respectively) and for the walking speed $\times$ target location (respectively $F(8, 144)=2.75$, $p=0.03$; $F(8, 128)=10.9$, $p<0.001$). A main effect for walking speed was found only for the target error ($F(2, 32)=8.1$, $p=0.001$).

Significant differences in *direction* (aggregated across *speed*) are denoted by braces in Fig. 3. The success rates of the 'narrower' and 'longer-narrower' cues are significantly lower than for 'wider', 'longer-wider', and 'longer', indicating that narrower steps are less likely to be performed correctly. The stepping errors have greater variability and follow an approximately inverse trend compared to the success rate; errors of the 'longer', 'longer-narrower', and 'narrower' cues are significantly larger than those of the 'wider' cue. Comparisons across *speed*, stratified by *direction*, show that walking speed significantly affects cued stepping performance during only the 'longer-narrower' and 'narrower' cues.

Due to the substantial challenge of the task for members of the amputee group, A1 and A2 performed the experiment at only the 0.4 m/s walking speed. The amputee group showed similar success rates as the control group for the 'wider', 'longer-wider', and 'longer' target locations. Strong differences between the amputee and control group became clear for the 'longer-narrower' condition, where the success rate was lower than the 95% quantile for both legs of A1 and A2 and the sound leg of A3.

3.2. Proactive adaptations

Fig. 4 shows the non-dimensionalised spatiotemporal gait parameters, their within-subject standard deviations (σ), and their symmetry indices (SI) for uncued walking in both the baseline and intermittent-cue sessions.

As seen in the left-hand panels, no significant anticipation (*session*) effects were found for step width. For step length (central panels), a significant *session* effect ($F(1, 18)=6.4$, $p=0.021$) but no significant *session* $\times$ *speed* effects were found. Post-hoc testing revealed that during walking speed of 0.8 m/s, the step length was significantly larger during intermittently cued walking than during uncued baseline walking. As shown in the right-hand panels, for step time, no significant *session* or *trial* $\times$ *speed* effect was found. A slight decrease in step length asymmetry was observed during the intermittent session at 0.6 m/s.

Pronounced differences were observed in within-subject standard deviations of these parameters. Main effects for *session* were found for step width variability ($F(1, 18)=44.1$, $p<0.001$), step length variability ($F(1, 18)=19.6$, $p<0.001$), and step time variability ($F(1, 18)=25.3$, $p<0.001$), and *session* $\times$ *speed* interaction effects for step width variability ($F(2, 36)=5.59$, $p=0.008$). Post-hoc comparisons showed significant increases in step width variability and decreases in step length variability and step time variability across all speeds from the 'uncued' session to 'intermittent'.

The amputee group did not exhibit a consistent trend of proactive adaptation. A1 showed little change across sessions, while A2 showed considerable increases in step length, step time, step width variability, and step length variability, and A3 showed an increase in step width variability and a decrease in step length asymmetry at the lowest walking speed.

3.3. Reactive adaptations

Fig. 5 shows the responses of the subjects to the cues while walking at 0.4 m/s, including the initial uncued steps (U), the cued target step (T), and the subsequent three response steps (R₁–R₃). For the 'wider' cue, the first response (R₁) step is wider than U and similar to T for the control group, then returns to the nominal width by R₂. Step length is slightly shortened in the R₂ and R₃ steps, while step time is significantly lower for R₁.

In the 'longer-wider' cue, both the R₁ and R₂ steps are significantly wider than the unperturbed steps. For the R₂ step, however, this difference is small. Stepping onto the target (T) results in increased step length, followed by a decreased step length in R₁. The step time of the R₁ and R₂ steps are shorter and longer than during unperturbed walking, respectively.

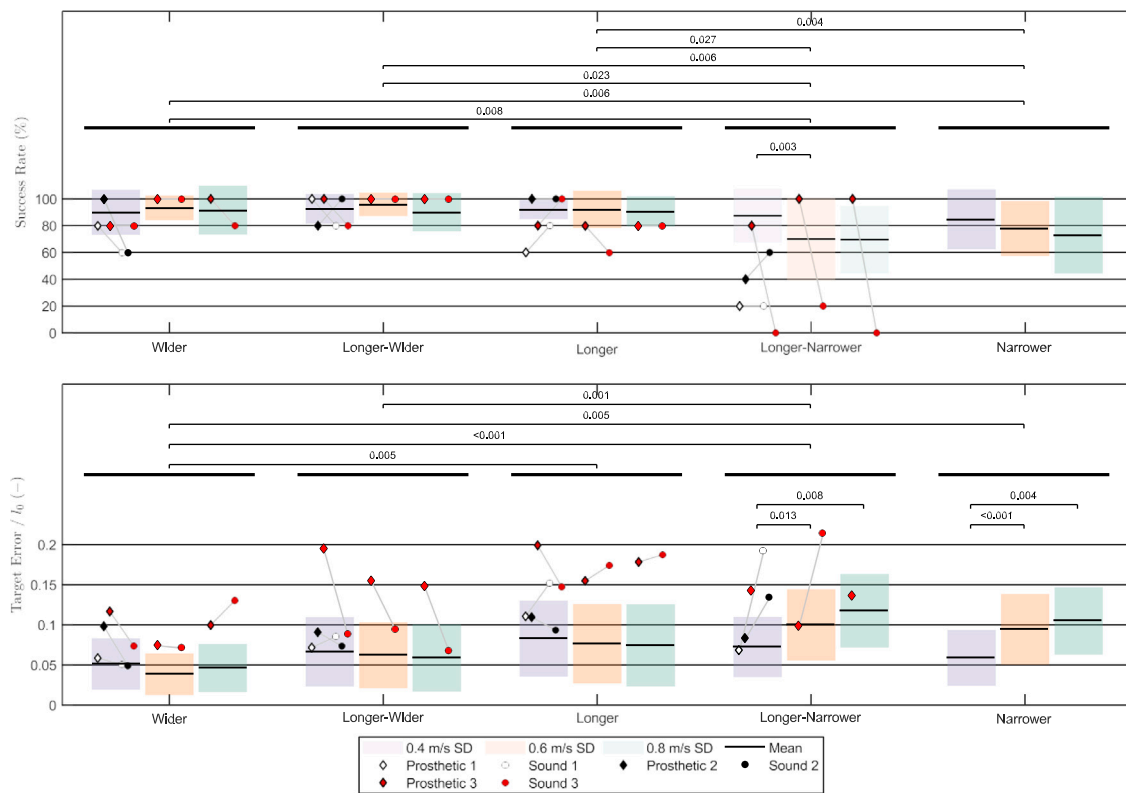


Fig. 3. Aggregated left and right data showing the success rate and error distance for each of the stepping cue locations and each of the treadmill speeds. Braces indicate significant ($p < 0.05$) post-hoc comparisons made across the factor *direction*.

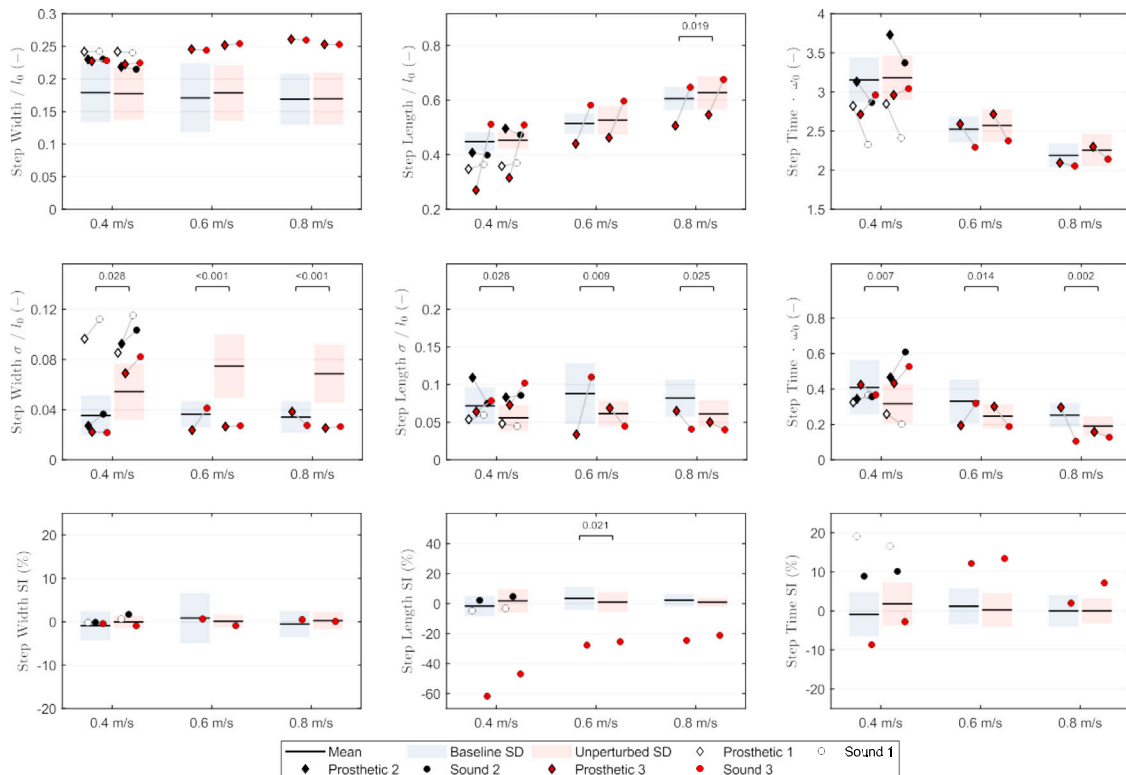


Fig. 4. Gait parameters and their within-subject standard deviations (σ) and symmetry indices (SI) for pre-experiment unperturbed walking ('baseline') and mid-experiment unperturbed walking ('unperturbed') sessions. The healthy control group ($n=19$) is depicted by boxes (the thick line is the group mean) of aggregated left/right data, and prosthesis users ($n=3$) by distinct markers for each subject and stepping limb. Braces indicate significant ($p < 0.05$) post-hoc comparisons made across factor 'session' of the healthy control group.

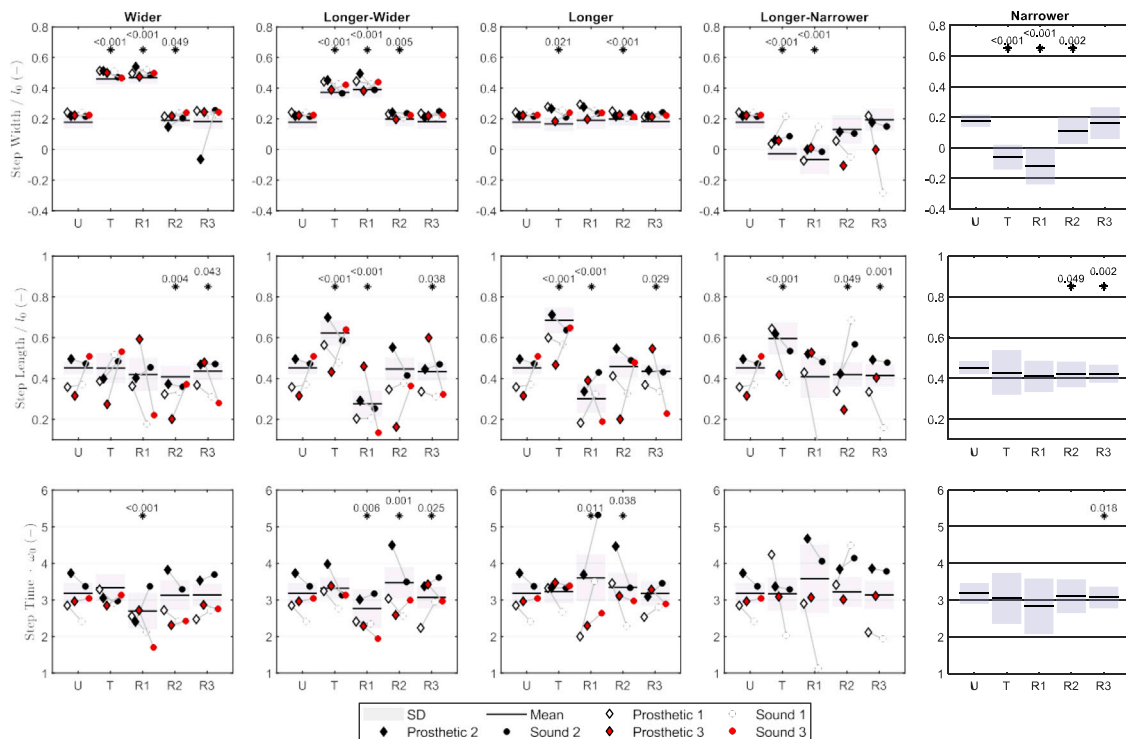


Fig. 5. Response to the self-induced perturbations while walking at 0.4 m/s showing uncued steps (U), the target step (T), and the subsequent response steps (R₁–R₃). An asterisk (*) and *p*-value indicates that the step is significantly (*p* < 0.05) different from U. Marker shapes of the response steps (R₁–R₃) indicate the limb used to make step T.

In response to the ‘longer’ cue, the step width is not substantially affected, except for a small increase in the R₂ step. The step length shows a similar response as to the ‘longer-wider’ target step: it is larger than U in the T step and smaller in the R₁ step; step time increases slightly in R₁.

For the ‘longer-narrower’ cue, the subsequent R₁ step is also narrower than the unperturbed steps. The negative step width of T and R₁ means that both the target step and the subsequent response step were cross-steps. The step length of the T step was longer, but the R₁ step was not significantly shorter, as was the case for the ‘longer-wider’ and ‘longer’ targets. Both R₂ and R₃ were significantly smaller than unperturbed walking. The step time was not substantially affected.

In response to the ‘narrower’ cue, the R₁ step was narrower than during unperturbed walking, and a cross-step for most of the control group. The R₂ step was still significantly narrower than during unperturbed walking but had a positive step width again for most of the group. The step width returned back to its unperturbed value in the R₃. As in all the other target directions, a slight persistent decrease in step length decreased was observed. The step time was not substantially affected by the narrower target stepping.

Owing to limited data from the amputee group, only the results from 0.4 m/s walking are discussed. The stepping responses of the amputee group followed trends similar to the control group, and most stepping parameters fell within or near the 95% quantile range for all stepping directions. The between-group differences observed in the unperturbed walking, such as wider steps and asymmetric step length for the prosthesis users, also persisted in the perturbed walking.

Notable differences between the control and amputee groups include a magnified asymmetry in the initial response steps (particularly in step length and step time) and a tendency to avoid large changes in foot placement (shorter step length in ‘longer’ and ‘longer-wider’ and wider step width in ‘longer-narrower’). In the individual responses, A1 and A3 were able to step on the ‘longer-narrower’ target only with their prosthetic leg, while A3 performed two more cross-steps during balance recovery than the average control subject.

4. Discussion

A task-specific platform and procedure for assessing the control of adaptive balance is presented with the intention of complementing existing balance assessment procedures. Unlike more conventional assessment approaches based on predictable stimuli and stereotype responses, the method presented here has more real-world challenge and relevance to tasks in daily living such as trip prevention and obstacle avoidance.

Cues prescribing narrower steps elicited the largest foot placement errors and multiple subsequent steps before returning to nominal gait patterns, together implying difficulty in achieving the adaptation and generally poor control authority in initiating and arresting this movement. Imposing narrower steps reduces the size of the base of support and may make maintaining the centre of mass securely inside this area more difficult [38]. Furthermore, the narrower steps reduce the mediolateral margins of stability [39], and thus impose a greater risk of collisions between the stance and swing legs [40], hence making balance control during gait inherently more challenging.

Although proactive adaptations of the gait parameters were not significant, exposure to adaptation cues yielded significant differences in gait parameter variability at all walking speeds — step width variability increased and both step length and step time variability decreased. Since the modulation of foot placement is a dominant control strategy in regulating the motion of the centre of mass during gait [41], changes in the variability of the spatiotemporal gait parameters may imply a change of control strategy in response to the presented movement task. An increase in step width modulation and corresponding decrease in step length and step time modulation may indicate that the movement task either requires a higher degree of control over lateral movement [42] or the attentional requirements of monitoring ground-level cues inhibit the use of vision in correcting unstable lateral dynamics.

This study shows that cued stepping, in addition to its application to studying gait and neurological processes, can be used to discriminate between pathological and healthy gait. Target hit rate is the clearest

discriminator between healthy and impaired gait, with differences being greater for narrower steps. Hence, gait with visual cues can augment other indices used for gait assessment such as the wider steps and greater asymmetry seen in the unperturbed walking of this study and other research [43,44]. Furthermore, although the stepping cues employed here represent real-world challenges, they do necessitate additional safety considerations.

The current setup has a number of technical limitations, and addressing these would lead to a more comprehensive assessment of balance capacity. For example, the platform would be enhanced through additional measurements and estimation such as the centre of mass, through which the aforementioned metric of the margin of stability could be computed and incorporated into the overall assessment of balance during gait. In addition, the present study is limited by the small sample size and incomplete condition coverage within the amputee cohort. The purpose of including the amputee cohort was to provide an initial clinical demonstration of discriminatory capability. Accordingly, the amputee results should be interpreted as exploratory and descriptive relative to the normative dataset. Furthermore, detailed prosthesis characteristics such as device mass were not recorded in this study. Prosthesis mass may influence gait dynamics and balance control but this factor could not be accounted for in the present analysis and should be considered in future work.

A further limitation of the platform is the use of a treadmill; it has been established that there key differences between treadmill and overground walking including angular momentum [45], foot placement [46], and in relation to kinetic and kinematic variables [47]. This may limit the extent to which gait assessment results from the BeStABLE can be applied to overground and everyday walking, as well as its utility as a gait training platform.

Future work will explore the boundaries of the capacity and willingness to self-perturb by varying the timing and magnitude of prescribed adaptations and investigating strategies to increase motivation during the gait training. Additionally, this procedure will be more extensively validated, including longitudinally tracking patients throughout the rehabilitation process, allowing the potential for a parallel therapeutic use of the system for mobility and balance training to be explored.

5. Conclusions

The results show that the BeStABLE, a treadmill-based augmented-reality environment displaying stepping targets intermittently at random moments, indeed enables discrimination between unimpaired control subjects and highly-functional below-knee amputees. Moreover, owing to its low effort for setup and short measurement times, the BeStABLE provides an efficient platform for standardised assessment. The study also highlights that the participants were unable or unwilling to follow certain cues — realising the requested narrower steps was too difficult for the participating prosthesis users. Hence, it is suggested to mainly use the longer-narrower condition to assess gait and balance impairment with the BeStABLE when time is restricted.

CRedit authorship contribution statement

Andrew Berry: Writing – review & editing, Writing – original draft, Methodology, Investigation, Conceptualization. **Nathan Fopma:** Methodology, Investigation, Conceptualization. **Matjaž Zadavec:** Methodology, Investigation, Conceptualization. **Andrew Pennycott:** Writing – review & editing, Writing – original draft, Methodology. **Heike Vallery:** Writing – original draft, Supervision, Investigation, Funding acquisition, Conceptualization. **Zlatko Matjačić:** Writing – original draft, Supervision, Funding acquisition, Conceptualization.

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Declaration of competing interest

The authors report no conflicts of interest. Furthermore, the authors alone are responsible for the content and writing of this article.

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