Evaluation of a Vibrotactile Biofeedback System Targeting Stance Time Symmetry Ratio of Individuals With Lower-Limb Amputation: A Pilot Study

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***Abstract*—Individuals with lower-limb amputation (LLA) often exhibit atypical gait patterns and asymmetries. These patterns can be corrected using biofeedback (BFB). Real-time BFB strategies have demonstrated to be effective to various degrees in BFB systems. However, no studies have evaluated the use of corrective vibrotactile BFB strategies to improve temporal gait symmetry of LLA. The aim of this study was to evaluate a wearable vibrotactile BFB system to improve stance time symmetry ratio (STSR) of LLA, and compare two corrective BFB strategies that activate either one or two vibrating motors at two different frequency and amplitude levels, based on a pre-set STSR target. Gait patterns of five unilateral LLA were assessed with and without BFB. Spatiotemporal and kinematic gait parameters were measured and assessed using a wearable motion capture system. Usability and workload were assessed using the System Usability Scale and NASA Task Load Index questionnaires, respectively. Results showed that participants significantly ($p<0.001$) improved STSR with BFB; however, this coincided with a reduction in gait speed and cadence compared to walking without feedback. Knee and hip flexion angles improved and changes in other parameters were variable. Immediate post-test retention effects were observed, suggesting that gait changes due to BFB were preserved for at least a short-time after feedback was withdrawn. System usability was found to be acceptable while using BFB. The outcomes of this study provide new insights into the development and implementation of clinically practical and viable BFB system. Future work should focus on assessing the long-term use and retention effects of BFB outside controlled-laboratory conditions.**

***Index Terms*—Biofeedback, gait symmetry, kinematics, lower-limb amputation, motor learning, rehabilitation, retention effects, spatiotemporal parameters, vibrotactile feedback, wearable systems.**

I. Introduction

LOWER-LIMB amputation is a major physical disability affecting over 57.7 million people worldwide [1]. With physical rehabilitation and the provision of a prosthesis, individuals with lower limb amputations (LLA) are typically able to regain their ability to walk; however, lifelong locomotor impairments often include reduced gait speed and various gait deviations [2]. One very common gait deviation seen in LLA is asymmetry, which is the degree of inequality of biomechanical parameters between limbs during a gait cycle [3]. Temporal gait asymmetry as is most commonly seen in unilateral LLA gait, is a result of spending more time weight-bearing on the intact limb compared to the prosthetic limb during walking [2], [3]. This appears to be a compensatory mechanism for the fact that the intact limb has a greater ability in maintaining balance compared to the prosthetic limb, especially in the early stages of the rehabilitation process [4]. However, temporal asymmetries can also be related to other factors such as the performance of prosthetic components and the physical abilities of the individual [5]. Ongoing rehabilitation including outpatient physiotherapy and gait training aim to improve an LLA's physical performance by targeting gait problems (such as asymmetry) to not only improve short-term outcomes (safer and more efficient gait) [6], but also to avoid long-term complications such as osteoarthritis due to overuse of the intact limb [7].

Technology-driven physiotherapy and gait training approaches such as virtual reality, therapy-focused videogames, and wearable biofeedback (BFB) systems can augment rehabilitation practices and improve the gait of individuals with mobility challenges, including LLA [8], Parkinso's disease patients [9], and stroke survivors [9]. A major benefit of wearable BFB systems is the ability to provide real-time feedback to reinforce physiotherapy goals and good gait habits [8], [10]. BFB-based gait training approaches consist of measuring specific gait parameters using sensors (e.g., pressure sensors, inertial sensors, goniometers) to provide real-time cueing (i.e., visual, auditory, haptic, or multimodal feedback) about the biomechanical status of the

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measured gait pattern or parameter [8], [9], [11]. Compared to visual or auditory, haptic feedback has been shown to be more suitable for field and community-based applications as stimuli perception is less prone to be affected by external conditions such as noise or visual distractions [8], [9].

Different strategies (i.e., coding schemes used to communicate biomechanical information to BFB users) have been shown to be effective to various degrees in BFB systems [12], [13], [14], [15]. Real-time BFB strategies can be categorized based on when the sensory stimulation is activated during the gait training session. Two commonly studied strategies include instructive and corrective feedback [14]. With instructive BFB, feedback is provided to the individual about the status of a specific gait parameter [13], [16]. Whereas, with corrective BFB, feedback is *only* provided when the specific gait parameter does not meet a specific target or range [12], [15], [17]. For example, if a patient drags their foot during the swing phase of gait, corrective BFB is provided if there is a chance of toe drag, measured for example by hip and knee flexion angles that are below a certain threshold. Whereas instructive BFB could be provided every gait cycle to remind or instruct the patient to increase their hip and knee flexion angles to prevent foot drag [14]. Martini et al. used instructive BFB delivering short-lasting vibrations around both sides of the waist, which was synchronized to the heel-strikes of each gait cycle to improve temporal symmetry of three unilateral transfemoral amputees (TFA) [16]. Findings showed that TFA increased their temporal symmetry, however, their gait speed decreased by nearly 50%. On the other hand, Plauche et al. compared corrective and instructive BFB by providing vibrotactile stimulation along the circumference of the thigh related to the center of pressure of the prosthetic foot to improve the sensory information about the prosthesis' position on the ground [17]. Results showed that non-disable participants walking with a prosthetic adapter improved their gait by reducing variability in their stride length and step width during both strategies; however, corrective feedback produced greater improvements for trunk sway variability [17]. The literature also suggests that corrective BFB may be preferred over instructive BFB specifically when delivering vibrotactile stimulation, as it reduces the possibility of the user becoming desensitized to the vibrations over time [17], [18]. However, to the authors knowledge, no studies have evaluated the use of corrective BFB to improve temporal symmetry of LLA gait.

Furthermore, while BFB systems typically target one specific gait parameter at a time (in this case temporal gait symmetry), it remains an important goal of rehabilitation-based gait training to improve overall gait patterns [3]. Many BFB studies have thus focused on investigating secondary changes in gait patterns [16], [17], [19], [20]. For example, Darter et al. used a real-time visual reality BFB system to assess spatiotemporal (walking speed, step length, stance time, step width) and kinematics (frontal-plane of motion for the trunk, pelvis, and hip angles) gait parameters of a TFA [19]. Results revealed improvements in trunk, pelvic and hip motion at the post-training session but not spatiotemporal parameters [19]. In the previously mentioned work of Martini et al., the improvements in symmetry came at the cost of slower walking speeds and undesired compensatory movement strategies at the trunk and pelvis [16]. These previous studies highlight the importance of not only studying the primary effects (i.e., targeted changes) but also other (i.e., spatiotemporal, kinematic, and kinetic) gait parameters related to locomotion.

The main objective of this study was to evaluate the use of corrective vibrotactile BFB to improve temporal gait symmetry of LLA when feedback is applied and immediately after, compared to no feedback. Two corrective strategies based on previous work involving non-disable individuals were investigated [12]; these were unilateral and bilateral stimulations as described in detail in the methods section. A secondary aim of this work was to evaluate the secondary (non-targeted) gait changes stemming from corrective BFB, including selected spatiotemporal and kinematic gait parameters. It was hypothesized that the proposed BFB strategies may improve the targeted gait parameter (stance-time symmetry ratio [STSR]), but simultaneously alter other gait parameters. The final aim was to assess the overall system usability and mental workload associated with the use of the BFB system; this is important for improving user interaction and informing the future development of clinically practical and viable BFB systems [13].

II. Methods and System Instrumentation

A. BFB System Instrumentation

The BFB prototype (Fig. 1b, 1c, 1d) was based on a previous iteration of the system [12]. However, since LLA often exhibit greater atypical gait patterns compared to non-disable individuals [21], the location and number of the force-sensitive resistor (FSR: model 406/402, Interlink Electronics, USA) sensors were modified (Fig 1d) to improve gait event detection (i.e., in cases of atypical foot contacts). Hence, the current system utilized fourteen FSR sensors (seven per foot) compared to eight FSR sensors [12]. Two tactors (model 307-103, Precision Microdrive, United Kingdom) provided real-time sensory feedback, via a microcontroller (Arduino Uno, Sparkfun Electronics, USA). A Bluetooth module (HC-05, Smart Prototyping, Hong-Kong) transmitted the information from the BFB system to a laptop for real-time data visualization, and a power bank (PowerCore 5000, Anker Innovations, China) powered the entire system.

B. Wearable Motion Capture System

Overall gait changes were assessed using a validated commercially available wearable motion capture system, Xsens MVN Awinda (Xsens North America Inc., USA), (Fig. 1a) to measure lower-limb kinematics (i.e., magnitude and timing of the ankle, knee, and hip joints, referred as joint angles for simplicity) and spatiotemporal parameters (i.e., gait speed, cadence, stride and step length) [22]. Seven inertial sensors were attached to the feet, lower leg (tibia, close to the knee) (x2), upper leg (middle of the lateral thigh) (x2), and sacrum (x1), (Fig. 1a). Joint angles and foot contact events were processed in the MVN Analyze software (Xsens North America Inc., CA, USA) [22]. Spatiotemporal and kinematic parameters were calculated as per ISB (International Society of Biomechanics) guidelines [23]. The symmetry ratio (SR) was used to quantify the changes in symmetry for stance time, step length and lower-limb joint angles (ankle, knee, and hip),

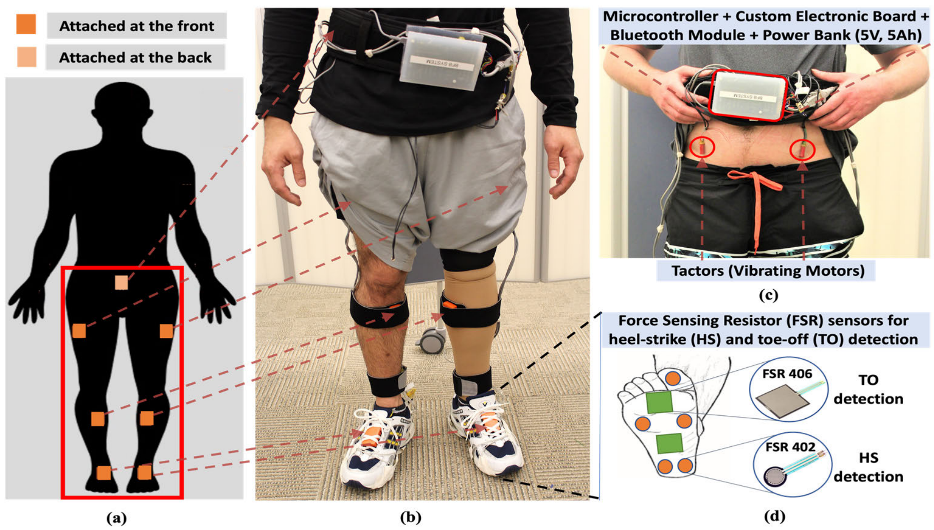

Fig. 1. a) A commercially available Xsens MVN Awinda system was used to acquire additional spatiotemporal and kinematics gait parameters, using seven inertial sensors attached to both lower limbs (feet, calves, thighs, and lower-back). b) Set-up of the Xsens and BFB system on a participant. The main BFB prototype components comprise of c) the Control Unit (microcontroller, custom electronic board, communication Bluetooth module, and power supply), and the Sensory Feedback Unit (two tactors) located at the lower abdomen at the prolongation axis of the rectus femoris muscle, and d) the Sensor/Transducer Unit consists of fourteen FSRs sensors (7 FSR per each shoe's sole), three underneath the heel and five underneath the toe (at the 1st and 5th lesser metatarsals, 2nd – 4th proximal phalanges, and the distal phalanx of the great toe). The black straps above the ankles were used to only hold the cables connecting the FSR to the microcontroller unit in place.

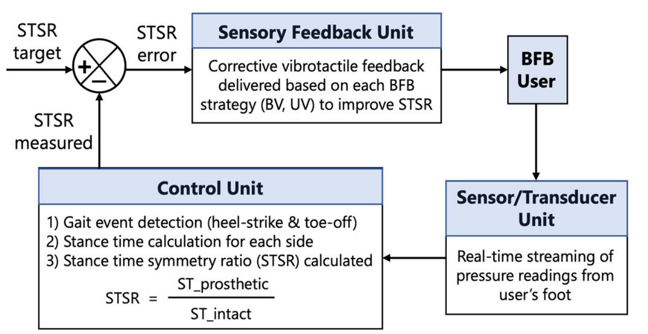

Fig. 2. BFB closed-loop control diagram targeting gait symmetry (STSR).

as per equation (1), where $X_{prosthetic}$ and X_{intact} represent the prosthetic and intact side, respectively.

$$Symmetry\ Ratio : SR = \frac{X_{prosthetic}}{X_{intact}} \quad (1)$$

C. BFB System Operation

The BFB system measured the real-time STSR and provided feedback targeting STSR (Fig. 2). STSR was calculated based on the stance time ratio of the prosthetic over the intact limb, as per equation (1), independently whether the prosthetic limb was the left or right limb. Stance time was defined as the amount of time that each limb remains in contact with the ground during a single gait cycle; timing difference between heel-strike (HS) and subsequent toe-off (TO) events of each foot [12]. The FSR threshold-based algorithm for HS and TO gait event detection computed the average maxima and minima of the FSR signals, as per [24]. Then, the FSR threshold levels were set midway between the calculated minimum and maximum FSR values to avoid susceptibility to possible spurious peaks at lower threshold levels and to minimize detection delays, which might occur at higher threshold levels [25]. STSR was measured for each stride, starting with HS of the prosthetic side. Once HS at the prosthetic side occurred, the previous step's stance time of both limbs (prosthetic and intact) was used to calculate the ratio. This value was then compared to the pre-set targeted STSR to determine if vibrotactile feedback should be provided at the beginning of the stance phase of the subsequent step. Vibrotactile feedback was based on the implemented corrective feedback strategies (i.e., a combination of vibration frequency/amplitude levels, vibration thresholds based on pre-set STSR thresholds, and the activation of one and two tactors).

D. BFB Strategies

The two unique corrective feedback strategies investigated were unidirectional control-variable vibration (UV) and bidirectional control-variable vibration (BV) (Fig. 3). These two vibrotactile strategies were previously found to be effective in altering temporal symmetry in non-disable individuals [12].

For UV, a single tactor (M1) was placed on the prosthetic side at the lower abdomen. For BV, two tactors were placed on the prosthetic (M1) and intact (M2) sides on the lower abdomen. UV provided feedback (M1 vibrates) only when the STSR was below the targeted STSR (Fig. 3a). BV provided feedback when the STSR was above (M2 vibrates) or below (M1 vibrates) the targeted STSR (Fig. 3b). For BV, only one tactor was activated during each gait cycle.

Tactors were activated at two different vibration levels. A larger difference between the actual and targeted STSR triggered a vibration at full (100%) power (frequency $\approx$ 250Hz, amplitude $\approx$ 7.5g), whereas a smaller difference produced a lower (50%) power vibration (frequency $\approx$ 125Hz, amplitude $\approx$ 2.8g) (Fig. 3). The UV strategy aims to encourage BFB users to increase their STSR until they reach and exceed the targeted STSR (Fig. 3a). Whereas the BV strategy requires the

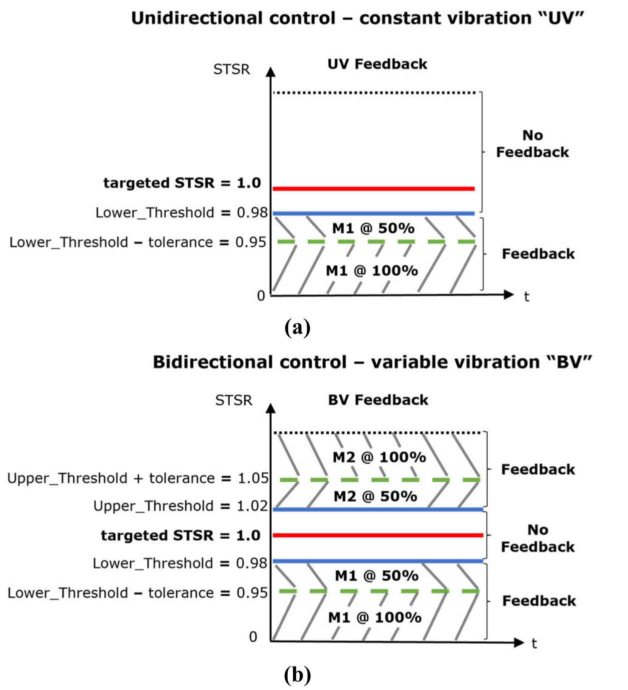

Fig. 3. Vibrotactile feedback strategies. (a) UV: Unidirectional control – variable vibration; (b) BV: Bidirectional control – variable vibration. UV uses a single tactor (M1) placed on the prosthetic side at the lower abdomen. BV uses two tactors placed on the prosthetic (M1) and intact (M2) sides at the lower abdomen. Tactors are activated at 50% and 100% of full intensity depending on difference between the actual and targeted STSR. The desired STSR value is denoted by targeted STSR. Vibration thresholds are denoted by Lower and Upper_Threshold, and by Lower and Upper_Threshold ± tolerance, respectively. Vibrations thresholds were selected based on typical STSR values for individuals with non-pathological gait (i.e., 0.95 ≤ STSR ≤ 1.05, where STSR = 1.0 denotes perfect gait symmetry).

user to maintain their STSR within a specified range (Fig. 3b). For both strategies, vibrotactile feedback was provided during the stance phase of the subsequent step, beginning at HS and ending at TO of the same limb within a particular gait cycle. Vibrations were only provided if participants walked with an STSR value outside of the pre-set STSR boundaries (Fig. 3).

The selection of the pre-set STSR boundaries for LLA were based on the normal STSR values for non-disable individuals, which are approximately equal to one (i.e., SR = 1.0 with 0.95 and 1.05 as the lower and upper CI boundaries, respectively) [26], [27], [28]. The testing paradigm for LLA for both feedback strategies (BV, UV) consisted of increasing participants' baseline STSR toward STSR = 1.0 (targeted STSR). Previous studies have reported baseline STSR values ranging from 0.70 – 1.0 for LLA, depending on the level of amputation and type of prosthesis [13], [16], [29]. STSR converging toward a value of 1 indicates gait symmetry improvement, whereas STSR diverging from 1 indicates asymmetry [3].

E. Participants

Five (n=5) unilateral LLA (1 female, 3 TFA and 2 TTA), age: 28.8 ± 6.8 years; height: 172 ± 5.4 cm; weight: 66.8 ± 7.4 kg, and time since amputation 21.1 ± 12.7 years were recruited (Table I). All participants wore a SACH foot. For TFA, P03 had a modular knee joint with rotary hydraulic (3R80, Ottobock, Germany), P04 had a modular brake knee with lock (Ottobock, Germany), and P05 had a knee rotationplasty. Participants had no known neurological disorders. The study was approved by the Research Ethics Board (REB #16-675) at Holland Bloorview Kids Rehabilitation Hospital, Canada. Informed written consent was obtained from each participant before commencing with the study.

TABLE I
PARTICIPANT DEMOGRAPHICS

Subject ID	Age (years)	Height (cm)	Weight (Kg)	Time since amputation (years)	Amputation level
P01	30	165	60	18	TTA
P02	29	172	72	23	TTA
P03	37	169	61	35	TFA
P04	30	175	77	28	TFA
P05	18	179	64	1.3	TFA

TTA: Transtibial amputee, TFA: Transfemoral amputee

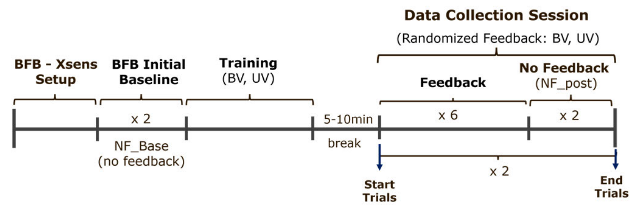

Fig. 4. Experimental protocol for the data collection session. Baseline trials consist of wearing the BFB system, but no feedback (NF) is provided. Feedback strategies (BV, UV) were randomized for data collection. BV: Bidirectional control – variable vibration. UV: Unidirectional control – variable vibration.

F. Experimental Protocol

Data were collected in a single session, as outlined in Fig. 4. Participants were instrumented with both the BFB prototype and wearable motion capture systems (Fig. 1). Both systems were then calibrated. For the Xsens, an N-pose calibration was performed [22] and for the BFB prototype system, two initial walk trials (NF_Base) were performed to establish the baseline STSR prior to receiving any feedback. This was then followed by a period of training where participants had the opportunity to walk using each feedback strategy (BV, UV).

During training, participants were coached about how to interpret the vibrotactile feedback. Verbal instructions and cues were provided such as "*if the motor on your prosthetic side vibrates, you need to spend more time in contact with the ground on that (prosthetic) side*". In terms of the vibration levels, cues included *"while walking, you will experience two different vibration intensities, the weaker vibration means you are closer to the target, and the stronger one means you are farther from the target. The goal is to receive no vibration*". Data collection for each participant comprised of 16 walking trials in total (6 feedback trials followed by 2 no-feedback trials for each of the two strategies) (Fig. 4). Each trial consisted of a 20-meter straight line walk at the self-selected gait speed. The order of the tested feedback strategies (BV, UV) was randomized with simple and balanced randomization using a random number generator.

G. Data Processing and Analysis

BFB data were captured using TeraTerm software (Tera Term Project, Japan), exported to a '.csv' file, and processed in

MATLAB (R2019b, Mathworks, USA) to extract the primary gait parameter (STSR) for each trial and feedback condition. Xsens data were exported to XML format using Xsens MVN Analyze software (Xsens North America Inc., USA). Data were then parsed using a python script which segmented the joint angles in the sagittal plane for each stride (HS to HS). For each stride, the maximum magnitude (joint angle in degrees) and corresponding timing (% of gait cycle) were extracted for the ankle (dorsiflexion, plantarflexion), knee (flexion, extension) and the hip (flexion, extension) joints. Average gait speed, cadence, stride length, and step length were also calculated. STSR was calculated based on the stance time ratio of the prosthetic versus the intact limb (Eq. 1). For other gait parameters (such as step length, ankle, knee, and hip joint angles), equation 1 was also used. The middle 8 gait cycles for each trial across feedback conditions for all the participants were used for data analysis. This ensured the same number of data samples for all participants and that parameters were acquired during steady-state gait.

Data were statistically analyzed in JMP Pro 2014 software (Statistical Discovery, SAS, USA). Shapiro–Wilk test ($p < 0.05$) for normality. Since all the parameters were non-normally distributed, a non-parametric two-tailed Wilcoxon signed-rank test was used for all spatiotemporal and kinematic data. Statistical significance was adjusted using a Bonferroni correction with a critical alpha level of 0.002 ($p = 0.05/21$) for all analyses to reduce potential type I errors. To assess immediate post-test retention effects on spatiotemporal and kinematic parameters, the no feedback conditions NF_Base and NF_post were compared within each strategy (BV, UV) using the two-tailed Wilcoxon signed-rank test.

H. Subjective Assessment of the BFB System

To quantify the subjective outcomes of BFB, the National Aeronautics and Space Administration Task Load Index (NASA-TLX) assessed the workload for each BFB strategy. Responses to the NASA-TLX questionnaire were captured, processed, and analyzed using the TLX mobile app (Official NASA Task Load Index (TLX) App, Version 10.3, 2016). The TLX app calculates the individual ratings and weights for each of the six subscales (mental demand, physical demand, temporal demand, frustration, effort, and performance) to obtain the overall score per feedback condition (BV, UV), as described by Hart et al. [30]. The NASA-TLX questionnaire was completed by participants immediately after the NF_post trials for each strategy (BV, UV).

The System Usability Scale (SUS), a 10-item questionnaire with a 5-item Likert scale, was used to quantify the overall usability of the BFB system (i.e., effectiveness, efficiency, and satisfaction) [31]. Scores were analyzed using the standard scoring method proposed by Brooke et al. [31]. According to Bangor et al., in terms of usability, a SUS score (scale 0 to 100) greater than a threshold of 50, 68, and 80 is considered “OK”, “acceptable”, and “good”, respectively [32]. The SUS questionnaire was completed at the end of the entire data collection session.

III. Results

A. Effect of BFB on Spatiotemporal Gait Parameters

For both TFA and TTA participants, significant improvements of STSR ($p < 0.001$) were found using both BFB strategies (BV, UV) compared to NF_Base (Table II). On average across all participants, STSR started at 0.89 ± 0.05 and increased by 4.6% to an STSR of 0.93 ± 0.05. For TFA, STSR started at 0.87 ± 0.04 and increased by 4.0% to 0.92 ± 0.05 and for TTA, STSR started at 0.91 ± 0.06 and increased by 5.1% to 0.96 ± 0.05. BV resulted in a greater improvement with an increase in STSR of 4.1% (TFA) and 6.2% (TTA), when compared to NF_Base (Table II). Whereas UV resulted in an increase in STSR of 3.8% (TFA) and 4.1% (TTA) compared to NF_Base. However, no significant differences were found between BV and UV for both groups. Further, participants exceeded the UpperTHR value (STSR value greater than 1.02) 17% and 10% of the time for UV and BV, respectively. This was calculated to further understand how the BV strategy maintained STSR values within the boundaries compared to UV.

Both groups (TFA, TTA) significantly reduced their gait speed ($p < 0.001$) and cadence ($p < 0.001$) during BFB compared to NF_Base (Table II). However, for TFA, gait speed and cadence decreased more during UV trials compared to BV, and vice versa for TTA (Table II). In addition, significant differences in gait speed and cadence ($p < 0.001$) were found between BV and UV for both groups (Table II).

Significant differences ($p < 0.001$) were also found for stride length, whereby UV resulted in the largest reduction for TFA and BV resulted in the largest reduction for TTA (Table II). Only TTA significantly decreased ($p < 0.001$) their step length symmetry during BFB compared to NF_Base (Table II). Specifically considering the prosthetic side, for both groups, step length of both groups significantly ($p < 0.001$) decreased during BV and UV compared to NF_Base (Table II). However, no significant differences were found between BV and UV for step length symmetry nor when comparing step length on the prosthetic and intact sides individually (except for step length of the intact side for TFA).

B. Effect of BFB on Lower-Limb Kinematic Gait Parameters

For TFA (Table III), compared to NF_Base, significant symmetry improvements were found in the magnitude of ankle dorsiflexion (BV, UV) and plantarflexion (BV, UV), but timing symmetry of ankle dorsiflexion (BV) and plantarflexion (BV, UV) worsened (Table III). For knee flexion, both magnitude (BV, UV) and timing (UV) symmetry improved. However, knee flexion timing symmetry (BV) worsened. Hip flexion, magnitude symmetry (UV) improved whereas timing symmetry (BV, UV) worsened. Comparing the two strategies to each other (BV and UV), significant differences were observed for knee flexion and hip extension timing symmetry.

For TTA (Table III), compared to NF_Base, ankle dorsiflexion magnitude symmetry (BV, UV) worsened, whereas plantarflexion magnitude symmetry (BV) improved (Table III). For knee flexion and extension, timing symmetry (BV) improved.

TABLE II
SPATIOTEMPORAL PARAMETERS

SPATIOTEMPORAL PARAMETER	MEAN ± STANDARD DEVIATION VALUES: NF_BASE	BV	UV	NF_POST BV	NF_POST UV	PAIRWISE COMPARISON (P-VALUES): NF_BASE - BV	NF_BASE - UV	BV - UV	RETENTION EFFECTS (P-VALUES): NF_BASE - NF_POSTBV	NF_BASE - NF_POSTUV
TRANSFEMORAL AMPUTEES (TFA)										
STANCE TIME SYMMETRY RATIO (STSR)	0.874 ± 0.040	0.910 ± 0.048	0.907 ± 0.057	0.937 ± 0.055	0.948 ± 0.074	**<.001**	**<.001**	0.0928	**<.001**	**<.001**
SPEED (M/S)	1.178 ± 0.071	1.035 ± 0.128	0.928 ± 0.147	1.130 ± 0.144	1.124 ± 0.094	**<.001**	**<.001**	**<.001**	0.543	0.019
CADENCE (STEPS/MIN)	97.615 ± 10.334	91.692 ± 2.164	82.982 ± 5.618	95.841 ± 5.088	95.118 ± 4.051	**<.001**	**<.001**	**<.001**	0.328	0.042
STRIDE LENGTH (M)	1.483 ± 0.055	1.431 ± 0.162	1.377 ± 0.166	1.495 ± 0.156	1.499 ± 0.144	0.0562	**<.001**	**<.001**	0.916	0.325
STEP LENGTH SYMMETRY	1.077 ± 0.174	0.919 ± 0.084	0.910 ± 0.150	0.912 ± 0.085	0.887 ± 0.127	0.0055	0.504	0.657	0.008	0.006
STEP LENGTH PROSTHETIC (M)	0.768 ± 0.059	0.685 ± 0.067	0.689 ± 0.104	0.711 ± 0.069	0.701 ± 0.075	**<.001**	**<.001**	0.880	**<.001**	**<.001**
STEP LENGTH INTACT (M)	0.723 ± 0.069	0.753 ± 0.108	0.726 ± 0.114	0.786 ± 0.101	0.803 ± 0.108	0.167	0.848	**<.001**	0.020	0.008
TRANSTIBIAL AMPUTEES (TTA)										
STANCE TIME SYMMETRY RATIO (STSR)	0.909 ± 0.060	0.965 ± 0.054	0.946 ± 0.055	0.958 ± 0.066	0.976 ± 0.050	**<.001**	**<.001**	0.0801	**<.001**	**<.001**
SPEED (M/S)	1.204 ± 0.120	0.852 ± 0.043	0.972 ± 0.037	1.143 ± 0.113	1.207 ± 0.143	**<.001**	**<.001**	**<.001**	**<.001**	0.506
CADENCE (STEPS/MIN)	115.990 ± 7.599	88.444 ± 6.282	99.725 ± 5.112	107.125 ± 6.468	114.766 ± 7.052	**<.001**	**<.001**	**<.001**	**<.001**	0.013
STRIDE LENGTH (M)	1.321 ± 0.055	1.229 ± 0.044	1.243 ± 0.038	1.349 ± 0.059	1.306 ± 0.114	**<.001**	**<.001**	0.1246	**<.001**	0.722
STEP LENGTH SYMMETRY	1.030 ± 0.128	0.938 ± 0.054	0.930 ± 0.032	0.932 ± 0.050	0.925 ± 0.053	**<.001**	**<.001**	0.132	**<.001**	**<.001**
STEP LENGTH PROSTHETIC (M)	0.669 ± 0.066	0.596 ± 0.020	0.602 ± 0.026	0.652 ± 0.044	0.640 ± 0.059	**<.001**	**<.001**	0.281	0.004	**<.001**
STEP LENGTH INTACT (M)	0.652 ± 0.025	0.637 ± 0.031	0.647 ± 0.019	0.699 ± 0.019	0.690 ± 0.028	0.003	0.539	0.127	**<.001**	**<.001**

Statistically significant differences (p < 0.002) are highlighted in bold font. (Adjusted critical alpha value = 0.05/21 = 0.002). Mean ± standard deviation. Initial baseline values (NF_Base) are color coded in light grey; followed by light and dark green for small and greater significant increments in relation to NF_Base values, respectively. Whereas small and greater decrements in relation to NF_Base values are color coded with light and dark orange, respectively.

Both, hip flexion magnitude symmetry (BV, UV) and extension timing symmetry (BV, UV) improved. Comparing the two strategies to each other (BV and UV), significant differences were observed for ankle plantarflexion and hip flexion magnitude symmetry, where increased improvements were observed with BV.

C. Immediate Post-Test Retention Effects

Examination of the data in (Table II and III) shows that many gait parameters that changed while corrective feedback was being provided remained changed once feedback was withdrawn. For spatiotemporal parameters, significant differences ($p < 0.001$) for both conditions (NF_postBV, NF_postUV) were found for STSR (in both groups), step length symmetry (only TTA), step length prosthetic (only TFA), and step length intact (only TTA) compared to NF_Base (Table II). Further, only TTA showed significant ($p < 0.001$) differences for gait speed, cadence, and stride length between NF_postBV and NF_Base, including step length prosthetic between NF_postUV and NF_Base (Table II). For kinematic parameters for the TFA group, significant symmetry improvements were found in ankle dorsiflexion magnitude symmetry (NF_postBV, NF_postUV), knee flexion timing (NF_postUV), and knee and hip extension timing (NF_postBV, NF_postUV) (Table III). However, symmetry worsened for ankle dorsiflexion timing (NF_postBV, NF_postUV) and knee flexion magnitude (NF_postUV) and timing (NF_postBV) symmetry. On the other hand, for the TTA group, significant symmetry improvements were found in ankle plantarflexion magnitude (NF_postUV), hip flexion (NF_postBV, NF_postUV) and extension (NF_postUV) magnitude. Symmetry worsened for ankle dorsiflexion magnitude (NF_postUV) (Table III).

D. Subjective Assessment of BFB System

Results of the NASA-TLX questionnaire (Table IV) showed that the overall workload (scale 0 – 100) for BV was 56.07 ± 8.07 and for UV was 53.00 ± 11.84, with frustration and mental demand as the lowest and highest factors, respectively.

TABLE III
KINEMATIC PARAMETERS

		MEAN ± STANDARD DEVIATION VALUES					PAIRWISE COMPARISON (P-VALUES)			RETENTION EFFECTS (P-VALUES)	
	KINEMATIC PARAMETER	NF_BASE	BV	UV	NF_POSTBV	NF_POSTUV	NF_BASE - BV	NF_BASE - UV	BV- UV	NF_BASE - NF_POST BV	NF_BASE - NF_POST _UV
	TRANSFEMORAL AMPUTEES (TFA)										
ANKLE	DORSIFLEXION (MAGNITUDE)	1.026 ± 0.016	1.017 ± 0.019	1.020 ± 0.014	1.018 ± 0.017	1.018 ± 0.014	**<.001**	**0.0021**	0.1271	**<.001**	**0.0023**
ANKLE	DORSIFLEXION (TIMING)	1.181 ± 0.254	1.271 ± 0.298	1.205 ± 0.279	1.245 ± 0.279	1.240 ± 0.289	**<.001**	0.0168	0.0628	**0.0015**	**<.001**
ANKLE	PLANTARFLEXION (MAGNITUDE)	1.195 ± 0.044	1.162 ± 0.079	1.142 ± 0.080	1.188 ± 0.077	1.194 ± 0.067	**<.001**	**<.001**	**<.001**	0.4633	0.6084
ANKLE	PLANTARFLEXION (TIMING)	1.141 ± 0.126	1.223 ± 0.068	1.246 ± 0.052	1.157 ± 0.125	1.177 ± 0.115	**<.001**	**<.001**	**<.001**	0.1271	0.0323
KNEE	FLEXION (MAGNITUDE)	0.997 ± 0.029	1.000 ± 0.034	1.003 ± 0.028	0.992 ± 0.023	0.995 ± 0.027	**<.001**	**<.001**	0.0168	0.0689	**<.001**
KNEE	FLEXION (TIMING)	1.087 ± 0.286	1.097 ± 0.290	1.076 ± 0.346	0.892 ± 0.311	0.988 ± 0.329	**<.001**	**<.001**	**<.001**	**<.001**	**0.0017**
KNEE	EXTENSION (MAGNITUDE)	1.024 ± 0.013	0.986 ± 0.026	0.978 ± 0.033	1.014 ± 0.041	1.008 ± 0.025	0.1585	0.0036	0.2526	0.0889	0.7700
KNEE	EXTENSION (TIMING)	0.979 ± 0.018	0.990 ± 0.017	1.000 ± 0.021	0.988 ± 0.017	0.986 ± 0.016	0.5426	0.6275	0.5887	**<.001**	**<.001**
HIP	FLEXION (MAGNITUDE)	1.025 ± 0.037	1.015 ± 0.058	1.006 ± 0.059	1.025 ± 0.053	1.028 ± 0.051	**<.001**	**<.001**	**<.001**	0.0751	0.7982
HIP	FLEXION (TIMING)	0.973 ± 0.016	0.988 ± 0.015	1.003 ± 0.016	0.984 ± 0.010	0.983 ± 0.014	0.0029	0.3295	0.2017	0.7147	0.0478
HIP	EXTENSION (MAGNITUDE)	0.967 ± 0.016	0.972 ± 0.019	0.976 ± 0.020	0.969 ± 0.019	0.966 ± 0.019	0.0414	**<.001**	**<.001**	0.9854	0.3203
HIP	EXTENSION (TIMING)	1.002 ± 0.028	0.988 ± 0.027	0.989 ± 0.032	0.998 ± 0.030	0.993 ± 0.027	**<.001**	**<.001**	**<.001**	**<.001**	**<.001**
	TRANSTIBIAL AMPUTEES (TTA)										
ANKLE	DORSIFLEXION (MAGNITUDE)	1.002 ± 0.015	0.988 ± 0.007	0.982 ± 0.010	0.997 ± 0.009	0.990 ± 0.016	**<.001**	**<.001**	0.0025	0.0211	**<.001**
ANKLE	DORSIFLEXION (TIMING)	1.005 ± 0.014	1.005 ± 0.015	1.007 ± 0.014	1.009 ± 0.012	1.002 ± 0.012	0.8051	0.9927	0.7981	0.0589	0.1944
ANKLE	PLANTARFLEXION (MAGNITUDE)	1.126 ± 0.067	1.097 ± 0.080	1.120 ± 0.098	1.121 ± 0.059	1.112 ± 0.077	**<.001**	0.5335	**<.001**	0.4087	**0.0011**
ANKLE	PLANTARFLEXION (TIMING)	1.117 ± 0.106	1.136 ± 0.126	1.143 ± 0.104	1.134 ± 0.110	1.144 ± 0.099	0.4632	0.6607	0.7421	0.1980	0.5444
KNEE	FLEXION (MAGNITUDE)	0.975 ± 0.012	0.984 ± 0.022	0.978 ± 0.019	0.973 ± 0.017	0.978 ± 0.010	0.0478	0.0357	0.6344	0.6214	0.5095
KNEE	FLEXION (TIMING)	0.846 ± 0.127	0.955 ± 0.251	0.907 ± 0.208	0.879 ± 0.119	0.866 ± 0.115	**<.001**	0.0168	0.0414	0.1912	0.7840
KNEE	EXTENSION (MAGNITUDE)	1.007 ± 0.039	0.997 ± 0.010	0.997 ± 0.014	1.004 ± 0.022	1.009 ± 0.029	0.0142	0.1951	0.1046	0.4087	0.1177
KNEE	EXTENSION (TIMING)	0.987 ± 0.010	0.997 ± 0.016	0.992 ± 0.013	0.990 ± 0.009	0.987 ± 0.011	**0.0018**	0.0235	0.0255	0.0033	0.1951
HIP	FLEXION (MAGNITUDE)	0.988 ± 0.013	0.995 ± 0.017	0.994 ± 0.015	0.993 ± 0.016	0.996 ± 0.014	**<.001**	**<.001**	**<.001**	**<.001**	**<.001**
HIP	FLEXION (TIMING)	0.989 ± 0.019	0.996 ± 0.019	1.000 ± 0.016	0.992 ± 0.014	0.991 ± 0.014	0.0223	0.5456	**<.001**	0.6808	0.6084
HIP	EXTENSION (MAGNITUDE)	0.993 ± 0.004	1.010 ± 0.008	1.004 ± 0.004	0.999 ± 0.009	1.001 ± 0.008	0.0036	0.0081	0.7011	0.0178	**<.001**
HIP	EXTENSION (TIMING)	0.989 ± 0.068	0.967 ± 0.037	0.992 ± 0.054	0.989 ± 0.048	0.995 ± 0.062	**0.0019**	**<.001**	0.4137	0.0818	0.6341

Statistically significant differences ($p < 0.002$) are highlighted in bold font. (Adjusted critical alpha value = 0.05/21 = 0.002). Mean ± standard deviation. Initial baseline values (NF_Base) are color coded in light grey; followed by light and dark green for small and greater significant increments in relation to NF_Base values, respectively. Whereas small and greater decrements in relation to NF_Base values are color coded with light and dark orange, respectively.

Based on the results reported by Grier et al., [33], our NASA-TLX scores are within the 50% (44.5) to 75% (66.5) of the global workload scores for a memory task (i.e., recall/recollection of stimuli) and within the 25% (50.9) to 50% (62.0) for a physical task (i.e., walking a designated route). Additionally, participants rated the usability of the BFB system as "acceptable" (mean score of 69.0 ± 7.8) based on the SUS and Bangor et al. [32].

TABLE IV
NASA-TLX SCORES FOR ALL PARTICIPANTS ACROSS FEEDBACK (WEIGHTED DIAGNOSTIC SUB-SCORES)

Factor	BV	UV
Mental Demand	345.0 ± 44.9	297.5 ± 90.9
Physical Demand	125.0 ± 75.0	132.5 ± 128.4
Temporal Demand	73.3 ± 61.9	125.0 ± 123.4
Performance	82.5 ± 67.6	48.8 ± 37.1
Effort	238.8 ± 173.1	131.3 ± 129.9
Frustration	47.5 ± 27.5	40.00 ± 20.0
Overall Weighted Score	**56.1 ± 8.1**	**53.0 ± 11.8**

IV. DISCUSSION

As hypothesized, all participants were able to significantly improve their stance time symmetry ratio (STSR) using corrective BFB strategies. Corrective vibrotactile feedback was provided only if BFB users were walking outside of the pre-set STSR thresholds, using two different vibration intensities for indicating relative closeness to the targeted STSR. Comparing the two strategies, BV produced slightly higher symmetry compared to UV, especially for TTA participants. This may be due to BV encouraging participants to reach and maintain their STSR values within certain boundaries. Furthermore, according to the percentage of gait cycles that were greater than the UpperTHR, BV provided better control of the STSR compared to UV.

A recent study showed that mean STSR values for LLA can range considerably (from 0.78 ± 0.08 to 0.97 ± 0.03) depending on the level of amputation and type of prosthetic knee (e.g., mechanical versus microprocessor prosthetic knee) [29]. TFA typically present with higher asymmetry caused by greater underlying physiological differences. TFA may also be more limited by their prosthesis (e.g., controller performance of the knee joint), when trying to achieve better more symmetrical gait [5]. Both TTA and TFA subgroups presented with STSR below 1, but as expected for TFA, STSR values were lower (Table II). With feedback, STSR increased for both groups, and slightly more for TTA compared to TFA (i.e., 5.1 % vs 4.0%, respectively); but neither group reached full symmetry with TFA remaining more asymmetrical. A previous study using BFB to induce STSR asymmetries (targeted STSR $=$ 1.10) in non-disable persons found an increase of participants' mean STSR by 11% and 8% for UV and BV, respectively, compared to no feedback [12]. These findings suggest that BFB can produce significant STSR changes; however, LLA compared to non-disable individuals (and TFA compared to TTA) might encounter additional challenges in achieving higher STSR values due to their prosthetic components or other physical differences. Given that it may not be possible or clinically ideal for some LLA to reach STSR values of 1, future work should explore target STSR values that are below 1, and perhaps customized to the individual based on factors such as their level of amputation and starting STSR.

A major finding of this study was the significant reduction in gait speed and cadence, as temporal symmetry was improved with the corrective feedback. Current literature suggests that temporal symmetry is velocity-dependent, and tha temporal symmetry increases with increased gait speed [16], [34]. However, with the introduction of an external system such as a BFB, studies have found an increase in temporal symmetry with a reduction in gait speed [12], [16]. A recent study found gait speed reductions of 13% to 23% in ten non-disable participants when improving STSR via vibrotactile feedback compared to no feedback [12]. Another study even found that gait speed of three TFA was reduced by half of the initial baseline with BFB [16]. It is possible that these reductions in gait speed are an effect of the cognitive load associated with the BFB strategies [35]. This may be supported by the results obtained from the NASA-TLX questionnaire, with mental demand (thinking, attention decisions making) being the highest-rated factor while using BFB, amongst the other subscales. A systematic review investigating gait performance with and without performing concurrent cognitive task showed that gait speed is primarily affected when increasing cognitive load [36]. Another study assessing cognitive workload in non-disable subjects during symmetrical, asymmetrical, and dual-task walking found increases in physiological parameters (heart rate, breathing frequency), with decreased stance time and increased cadence during high cognitive load tasks [37]. Accordingly, it might be expected that after extended gait training periods, the proposed BFB strategies might have the potential to improve gait symmetry (spatiotemporal and kinematics) without significantly increasing the cognitive load as the process becomes more subconscious, which might allow individuals to maintain their preferred gait speed and cadence.

Additionally, future work should explore the incorporation of progressive BFB gait training strategies to reduce cognitive demands while learning and performing a new motor task such as BFB directed gait. This may include, as mentioned earlier, starting with a target STSR that is less than 1, and/or increasing the BFB thresholds to provide less stringent feedback initially in the gait training process. Progressive BFB gait training strategies allow for the acquisition of a new complex motor skills that can be gradually achieved based on a series of simple and less taxable activities that progressively promote motor learning (i.e., acquisition, consolidation, and retention) towards completely learning the new motor skill [38]. Acquiring a new motor skill involves cognitive, associative, and autonomous stages [38]. These stages are initially characterized by progression from conscious, slow, and inefficient movements towards more accurate, consistent, and efficient ones, suggesting that the motor skill has been learned [38].

Gait changes were found in several kinematic symmetries during both strategies (BV, UV). Effects on ankle joint angle symmetry were mixed, as all participants used a passive fixed-ankle prosthesis that restricted ankle range of motion and movement. However, improvements in knee flexion magnitude symmetry were observed in the TFA group and hip flexion in both groups using both strategies. For instance, Fig. 5 displays kinematic changes among conditions (NF_Base, BV, UV) for one TFA participant (P03). The overall knee flexion magnitude decreased on both limb sides due to the slower gait speed. However, the changes were more significant on the prosthetic

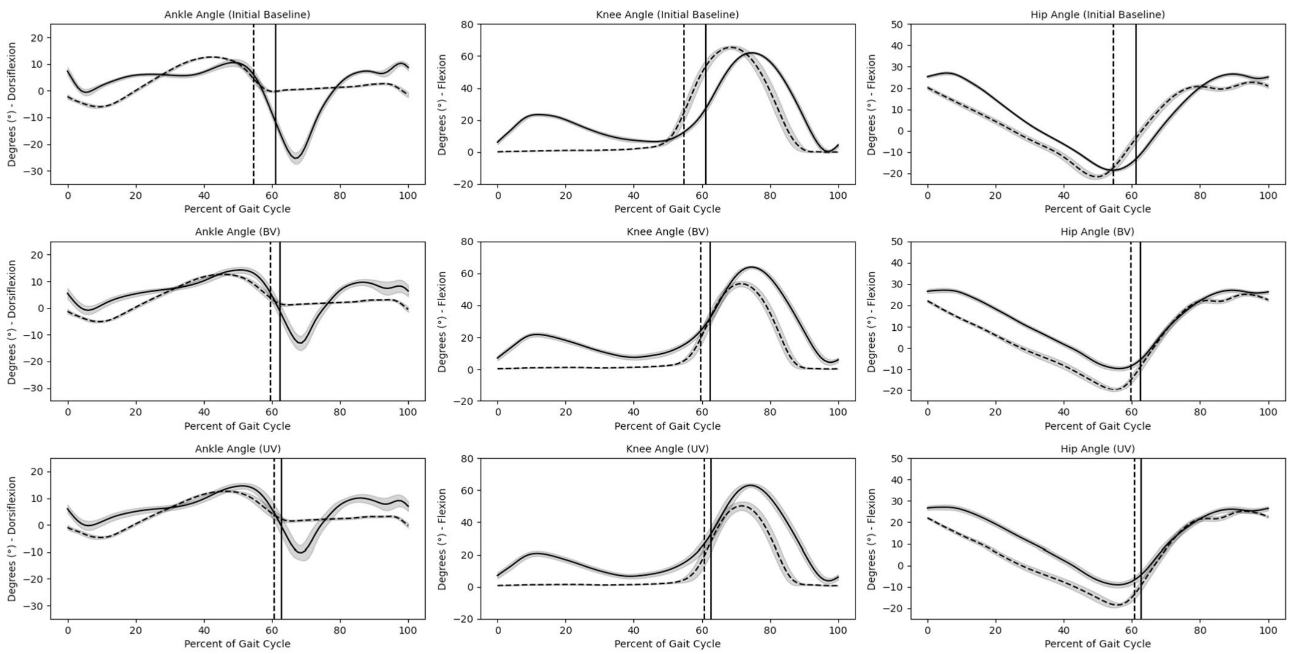

Fig. 5. Lower-limb kinematic parameters (ankle, knee, and hip joints) among conditions (NF_Base, BV, UV) for participant P03 (TFA). Solid line refers to prosthetic side and dashed line refers to intact side. Shading of each line indicates the corresponding standard deviation. Vertical lines indicate the toe-off (TO) time as % of gait cycle. TO times relate to the speed at which the participant is walking (more symmetrical during BV and UV compared to initial baseline).

side compared to the intact side, resulting in improved knee flexion symmetry. Hood et al.'s database comparing lower-limb kinematic data of 18 amputees, indicated that at slower gait speeds, reduction in knee flexion is more prominent on the prosthetic side compared to the intact side [39]. It is evident that kinematic changes for both sides do occur during the BFB-based walk trials, and therefore it is important to evaluate these secondary parameters and isolate potential confounding effects including those related to gait speed changes [40], [41].

Evaluating the immediate post-test retention effects, suggests that some immediate retention exists, as STSR and many other gait parameters that changed with feedback, remained changed with the withdrawal of feedback. Hence, these small improvements towards more symmetrical gait might suggest that significant and permanent gait changes can potentially be achieved gradually over longer durations of gait training.

Results from the SUS questionnaire indicated the overall usability of the BFB system as "acceptable", with similar results observed in other studies using BFB for LLA gait training [13]. Future work should focus on increasing usability of the system by considering more streamlined designs with wearable technologies (i.e., inertial sensors).

The limitations of this study include the small sample size which restricted generalizable conclusions for the spatiotemporal and kinematic gait parameters. Additionally, BFB strategies such as rhythmic stimulation [20], capable of controlling symmetry and walking speed/cadence concurrently, should be investigated. In terms of wearability and ease of use, microelectromechanical systems have advanced significantly over the past few decades, providing reliable movement measurements through devices such as inertial measurement units (IMUs) [42]. Accordingly, the use of IMUs can be an advantage over FSR sensors since they can be easily attached anywhere on the body and do not need to be maintained as frequently as FSR sensors. Thus, adapting the proposed BFB system to use IMUs would potentially increase its wearability and portability.

V. Conclusion

This paper evaluated a wearable vibrotactile BFB system using corrective BFB strategies to improve temporal gait symmetry. The system improved stance time symmetry ratio in unilateral LLA, but resulted in changes within other spatiotemporal and kinematic gait parameters. Future work should explore long-term BFB use which may lead to reduced cognitive demands and further improvements in overall gait.

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