A Haptic Feedback System for Phase-Based Sensory Restoration in Above-Knee Prosthetic Leg Users

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***Abstract*—Persons with amputations lack important senses from the amputated limb. With the absence of proprioception in the amputated leg, amputees have far more difficulty maintaining a natural gait with balance and stability. The biggest determinant of temporal limb behavior during locomotion is the phase in the gait cycle, which can be estimated using the center of pressure (COP) under the feet. We hypothesize that feedback from the COP of the prosthetic foot can help restore a more robust sense of phase in transfemoral (above-knee) amputees. This paper presents a device that provides vibrotactile feedback based on the COP from the prosthesis, providing proprioception and potentially an improved sense of phase to the user. Experiments showed that the haptic device significantly decreased variability of stride length, step width, and trunk sway in novice (able-bodied) users of a transfemoral prosthetic leg during treadmill locomotion (N=9), indicating improved gait stability.**

***Index Terms*—Haptic Feedback, Vibrotactile, Prosthesis, Center of Pressure, Amputees**

I. INTRODUCTION

LOWER-limb amputees, especially at the transfemoral level, exhibit a number of problems in their gait such as wider steps than able-bodied subjects [1], asymmetric walking gaits [1], increased variability in the medio-lateral acceleration of the trunk [2], inferior standing balance [3], and increased falling rates [4]. Many of these issues can be attributed to the lack of sensory feedback from the missing limb. This paper presents a prototype system that is intended to work with any commercial prosthesis and restore a key source of sensory feedback to the human user, specifically that related to the center of pressure (COP) under the prosthetic foot.

The proprioceptors in the limbs provide the central nervous system with information such as joint angles, muscle length, and muscle tension [5]. It is well known that vibrotactile feedback can activate these proprioceptors. For example, Craske et al. induced the perception of impossible limb positions by vibrating the limb's tendons [6]. Craske et al. used a linear actuating vibrator driven sinusoidally at 80 Hz on the tendons of the arm to create a sensation that the subject's arm was moving. They could "move" the arm beyond the normal physical limitations, with no pain to the subject. Another group found that tactile information on the plantar sole is used for perceptual purposes (i.e., proprioception) [7]. Roll et al. were able to create kinesthetic illusions by applying vibrations with a frequency of 100 Hz on the soles of their able-bodied subjects. The subjects felt as though they were tilting in the same direction of the vibrated plantar site [7].

The perceived location of the center of pressure (COP) on the plantar sole could be responsible for the kinesthetic illusions in [7]. The COP can be measured by mechanoreceptors in the planar sole, which are stimulated by vibrations at certain frequencies. These mechanoreceptors are known to contribute to certain aspects of postural control during human walking [8]. The heel-to-toe movement of the COP during locomotion could provide sensory feedback indicating the phase of gait [9], [10]. In a healthy subject, the COP follows a regular path along the plantar sole, moving from the heel to toe, which can be mapped to a corresponding position in the gait cycle. The path of the COP is also a commonly used variable in clinical practice for quantifying dynamic balance [11].

An amputee may be disadvantaged without sensory feedback related to the COP of the prosthetic foot. The simulation model from [12] suggests that the COP path on the residual thigh of above-knee amputees does not correlate well with the COP path of the prosthetic foot. Therefore, socket pressure may not be enough for amputees to walk correctly on prosthetic legs. Even if an amputee were to obtain a prosthetic leg that could walk normally, he/she may not have a good sense of what the leg was doing without COP information. The prosthetic leg user would likely still have imbalance in gait, which could result in falls and/or a fear of falling [13]. Feedback related to the COP location under the prosthetic foot may be needed to improve stability in amputee locomotion.

The goal of this work is to investigate a haptic feedback system that is designed to give an amputee a sense of prosthesis position on the ground based on the COP of the prosthetic foot. We hypothesize that the lack of this COP feedback causes a diminished sense of gait cycle phase, which contributes to inferior gait in lower-limb amputees. We further hypothesize that restoring this feedback through a haptic device could reduce variability in kinematic measures associated with gait stability or balance (e.g., [14], [15], [16]). This system could aid an amputee in monitoring the motion of the prosthesis and providing a course of correction whenever needed.

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The idea of using a feedback system to improve postural stability and balance is well established [17], [18], [19]. Feedback related to heel/toe contact and the COP has been provided by visual or auditory means [20], [21], but vibrotactile sensory substitution systems have a higher acceptance rate than other sensory substitution systems [22]. Vibrotactile devices for balance commonly provide feedback based on trunk tilt [23], [24]. One study provided vibrotactile stimulation to the residual limb of transtibial amputees based on force measurements from the prosthetic foot, but no significant benefits to postural stability were observed during stationary standing experiments [25]. It is possible that the interaction forces between the residual tibia and the prosthetic foot provide sufficient feedback related to the COP, whereas this might not be the case for transfemoral amputees.

Our system is unique in that it is designed to improve *dynamic gait* in *transfemoral* amputees by geographically mapping the location of the prosthetic COP to vibrotactile feedback around the residual thigh. This paper presents the design and experimental validation of the haptic system with 9 novice (able-bodied) users of a transfemoral prosthetic leg during treadmill walking. Our findings indicate that the haptic device enabled a statistically significant decrease in the variability of step width, stride length, and trunk sway, indicating improved gait stability.

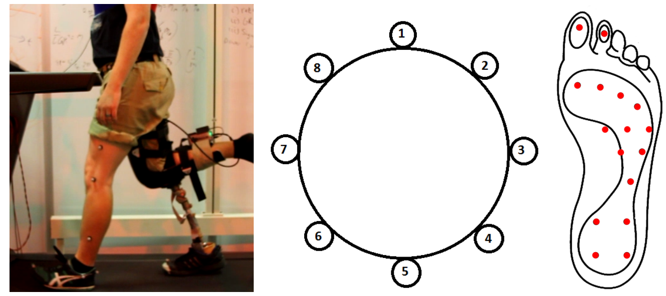
II. Methods

A. Subjects

Nine able-bodied (non-amputee) subjects participated in this study (six males, height 1.79 ± 0.1 m; weight 68.0 ± 9.4 kg; age 25.6 ± 2.0 years; mean ± standard deviation). Able-bodied subjects were chosen over amputees in order to have a control group to validate the stability measures in this feasibility study. Amputee subjects would also have required the design of custom, subject-specific sockets with embedded vibration tactors. This would require extensive work for each subject, which was beyond the scope of this feasibility study. Able-bodied subjects wore a bypass adapter to walk on a right-legged, above-knee prosthesis (Fig. 1a), simulating transfemoral amputee gait. The prosthesis used in our experiments included a solid ankle, cushioned heel prosthetic foot, and a passive single-axis constant friction knee. With this configuration, the prosthesis provided no active assistance to the users. All subjects were young, active, healthy adults. Informed consent was given in accordance with the University of Texas at Dallas Institutional Review Board (#15-43).

B. Haptic Feedback System

The haptic feedback system comprised eight linear resonant actuator (LRA) motors (C2 tactors: 1.2” diameter, 0.31” thickness, 17 grams, Engineering Acoustics, Inc., Casselberry, FL) that were inserted into eight small pockets made from scrap pieces of neoprene and sewn onto the top-most section of an elastic neoprene knee support wrap, approximately 3 mm thick. The tactors were positioned to be equally spaced along the circumference of the thigh, as illustrated in Fig. 1b, approximately 4 inches above the knee. The C2 tactors



(a) Experimental setup (b) Knee sleeve (c) Insole

Fig. 1. (a) An able-bodied subject walking on the prosthetic leg using a bypass adapter with the haptic device. (b) The positions of the vibrotactile motors (i.e., tactors) around the thigh, where 1 is on the front of the thigh. (c) The positions of the FSRs on the insole, indicated by red dots.

activated at 250 Hz, which has been shown to be ideal for activating Pacinian corpuscles in the skin, a key mechanoreceptor that responds to high frequency vibration stimuli [26]. The tactors were driven at approximately 0.25 A rms and 12 Vdc amplitude. The same amplitude was used for all subjects.

The pressure under the prosthetic foot was measured via a custom, 3D printed, rubber insole equipped with sixteen force sensing resistors (FSRs). The FSRs were positioned under the major pressure points of the foot (Fig. 1c) according to Shu et al. [27]. An Arduino microcontroller (MEGA 2560 R3: ATmega2560, 16 MHz, Arduino LLC, Boston, MA) measured the resistance of the FSRs, calculated the COP, and determined the appropriate feedback. The x-y coordinates of the COP were calculated using a weighted average formula,

$$COP(x,y) = \left(\frac{\sum (P_i x_i)}{\sum P_i}, \frac{\sum (P_i y_i)}{\sum P_i} \right), \tag{1}$$

where P_i is the pressure at the i^{th} FSR and x_i and y_i are the coordinates of the i^{th} FSR.

Two different feedback strategies were examined. The first, Constant Feedback (CNST), continuously informed the user of the location of the prosthesis COP via geographically mapped vibrating feedback during the prosthetic stance period. For example, if the COP was in the heel region, then the feedback would be on the back of the thigh. As the subject walks with the device, the feedback travels around the thigh to signify that the COP is moving forwards towards the toe region.

The second strategy, Corrective Feedback (CORR), provides vibrating feedback in the direction of error when the prosthesis COP significantly deviates from its nominal path. For example, if the subject’s COP is on the far left side of the insole, then the feedback vibrates the left side of the thigh. The nominal COP path is based on prior results by Han et al. [11]. If the COP is determined to be within approximately 1 cm of the nominal COP path, then the feedback stops. The CORR feedback was designed with the idea that the user might ignore or become desensitized to the feedback over time. Prolonged use with vibrotactile stimulation has been shown to cause desensitization of the mechanoreceptors in the skin [28].

If a lost sense of phase is indeed a contributing factor to inferior gait in lower-limb amputees, then either feedback strategy might have a beneficial effect. The CNST feedback is a direct mapping of the COP, which continuously provides feedback related to the gait cycle phase. Ideally, the CORR strategy provides feedback only when the COP trajectory is non-normative, i.e., when the user when the user is outside the normal starting point on the heel or ending point on the toe. It therefore gives the user an indirect sense of phase during gait. Experiments were designed to determine which one provides better results and to compare with no-feedback control trials.

C. *Procedure*

Motion capture information was collected with a ten-camera Vicon marker based system (T20S, Oxford, UK). Reflective markers were placed on sixteen locations on the lower body of the subject: the posterior superior iliac, anterior superior iliac, both left and right thighs, knees, shanks, ankles, heels, and above the second metatarsal toe head. These locations allowed for reconstruction of kinematic information of the legs. Marker data was collected and synchronized with Vicon Nexus 1.8.5 software for later analysis. For the experimental trials, subjects walked on a treadmill for approximately 30 seconds at a self-selected speed. We conducted 10 trials per condition per subject. This far exceeds the suggested number of strides for most gait stability measures [29]. After each experiment, the subject was given a questionnaire to assess their level of confidence and sense of stability.

The subject was informed of each testing condition and how to interpret it. There were four testing conditions: able-bodied control (CTRL), corrective feedback (CORR), constant feedback (CNST), and prosthesis control/no-feedback (NF). The able-bodied control trials have the subjects walking without the use of the prosthesis and without the feedback system, representing a ‘best-case’ scenario for gait stability. In the prosthesis control, the subjects walk on the treadmill without the use of the feedback device while using the prosthetic leg. This condition gave a ‘worst-case’ scenario, simulating the gait of a novice transfemoral amputee. The order of the trials were randomized for every subject to eliminate the effect of any learning curve with the prosthesis. If the subject took a misstep or fell while walking during a trial, that data was discarded and the trial was restarted. After all trials were conducted, data was post-processed using the Vicon Nexus ‘Plug-in Gait’ module. All further analyses were performed using Matlab R2013B (The MathWorks, Inc. Natick, MA).

The subject’s step width, stride length, and trunk sway were calculated for each step. The variances of these measurements are known to differentiate between subject groups such as young vs. elderly [14] and fallers vs. nonfallers [15]. An increased variability in transtibial amputees may indicate an increased chance of falling [16]. These measures were validated for our study by comparing the CTRL and NF trials before examining the effects of the haptic feedback.

The step width was calculated as the distance on the x-axis from the left ankle marker to the right ankle marker during double support stance. The stride length was calculated as the distance that the left heel marker traveled on the y-axis in one stride, corresponding to the sound leg during the prosthetic trials. Stride length and step width are both important as symmetry and regularity of walking are two important aspects in gait analysis, especially in transfemoral amputees [30]. The trunk sway was simply the trajectory of the right posterior superior iliac (PSI) marker on the x-axis. This provided insight into swaying medio-laterally at the hip. Medio-lateral sway has been used in many papers to effectively determine balance performance [31], [32], [33]. It has also been shown that medio-lateral trunk acceleration variability is higher in transfemoral amputees [2].

TABLE I
P-VALUES OF LEFT-TAILED F-TEST FOR CTRL VS NF

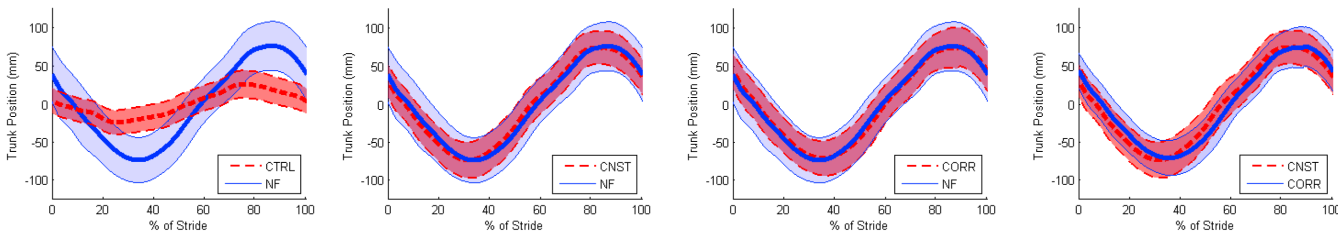
Subject	SLV	SWV	TSV
01	$\ll 0.05$	$\ll 0.05$	$\ll 0.05$
02	$\ll 0.05$	$\ll 0.05$	$\ll 0.05$
03	$\ll 0.05$	0.88	$\ll 0.05$
04	$\ll 0.05$	$\ll 0.05$	$\ll 0.05$
05	$\ll 0.05$	$\ll 0.05$	$\ll 0.05$
06	1	$\ll 0.05$	$\ll 0.05$
07	$\ll 0.05$	$\ll 0.05$	$\ll 0.05$
08	$\ll 0.05$	$\ll 0.05$	$\ll 0.05$
09	0.5	1	1
Lumped	$\ll 0.05$	$\ll 0.05$	$\ll 0.05$

For step width and step length we examined variability from step to step, whereas for trunk sway we examined the variance of the trunk trajectory about vertical. From here on the following abbreviations will be used: stride length variability (SLV), step width variability (SWV), and trunk sway variability (TSV). To compare variability between trial conditions, a F-test (Levene’s test) was used with a confidence level of $\alpha = 0.05$. Levene’s test was used because some of the data appeared to be non-normally distributed due to outlying values. The two feedback strategies were compared to each other and the NF control. The F-tests were conducted on variables SLV, SWV, and TSV for each subject. For all comparisons, stability was considered ‘improved’ if the P-value from the F-Test was less than 0.05. This means that the F-test concluded that the variabilities of the two datasets are different (one less than the other) with at least 95% confidence.

III. RESULTS

A. *Control vs No-Feedback*

The purpose of this comparison was to demonstrate the sensitivity of SLV, SWV, and TSV as stability measures for our study. The CTRL cases are expected to have the smallest variance out of all trials conducted. Almost all of the subjects showed a significantly larger variance in SLV, SWV, and TSV for the NF case compared to CTRL (F-test, $P<0.05$). Two outlier cases were present in the data. One subject showed a smaller variance in SLV for the NF case, and another subject showed a smaller variance for SWV and TSV. Table I shows the P-values for the subject-specific comparison. In Fig. 2a, the trunk sway under the CTRL testing condition shows significantly smaller standard deviation than the NF condition. The overall range of the NF condition is about 200 mm, compared to the CTRL condition which shows an overall range of 50 mm. This means that the subject’s trunk did not move as far in the CTRL condition as in the NF condition.



(a) Control vs No Feedback (b) Constant vs No Feedback (c) Corrective vs No Feedback (d) Corrective vs Constant Feedback

Fig. 2. Temporal trunk sway data for a representative subject. The thicker center line represents the average trunk sway trajectory, whereas the light colored areas represent the ($\pm$) standard deviation. These figures represent data from one subject across all trials. The data's time period was normalized to be the same and the data was shifted up/down on the y-axis to have a mean value of zero.

TABLE II
P-VALUES OF LEFT-TAILED F-TEST FOR CNST VS NF

Subject	SLV	SWV	TSV
01	$\ll 0.05$	$\ll 0.05$	1
02	0.21	1	$\ll 0.05$
03	0.94	$\ll 0.05$	1
04	$\ll 0.05$	$\ll 0.05$	$\ll 0.05$
05	$\ll 0.05$	$\ll 0.05$	$\ll 0.05$
06	$\ll 0.05$	$\ll 0.05$	0.18
07	$\ll 0.05$	$\ll 0.05$	1
08	$\ll 0.05$	$\ll 0.05$	$\ll 0.05$
09	1	1	1
Lumped	$\ll 0.05$	$\ll 0.05$	1

TABLE III
P-VALUES OF LEFT-TAILED F-TEST FOR CORR VS NF

Subject	SLV	SWV	TSV
01	0.99	$\ll 0.05$	1
02	$\ll 0.05$	$\ll 0.05$	$\ll 0.05$
03	0.12	0.078	$\ll 0.05$
04	$\ll 0.05$	$\ll 0.05$	$\ll 0.05$
05	$\ll 0.05$	$\ll 0.05$	$\ll 0.05$
06	$\ll 0.05$	$\ll 0.05$	$\ll 0.05$
07	0.09	$\ll 0.05$	1
08	$\ll 0.05$	$\ll 0.05$	$\ll 0.05$
09	1	$\ll 0.05$	$\ll 0.05$
Lumped	$\ll 0.05$	0.45	$\ll 0.05$

B. Constant & Corrective Feedback vs No-Feedback

The variances in the CORR and CNST feedback cases are expected to be smaller than the NF case. The P-values for these two comparisons are in Table II and Table III. For stride length, six subjects showed significant improvement, i.e., reduced variance, using the CNST feedback strategy and five subjects showed significant improvement using the CORR feedback strategy. All other subjects showed no significant change. For step width, seven subjects showed a significant decrease in variance using the CNST feedback strategy. The other subjects showed no significant change. For the CORR feedback strategy, eight subjects showed a decrease in variance and the other subject showed no significant change. For trunk sway, four subjects showed significant decrease in variance while using the CNST feedback strategy, whereas five subjects showed an significant increase or no significant change. Seven subjects showed a significant decrease in variance using the CORR feedback strategy and two subjects showed no change.

Figs. 2b and 2c show the trunk sway temporal data for the two feedback strategies compared to the NF condition. An improvement in variability across the entire stride is visible from the narrower error bars for either feedback strategies compared with the NF condition. The mean value for the trunk sway temporal data did not visually change, but only variability is associated with stability [29].

C. Corrective Feedback vs Constant Feedback

The purpose of this test was to determine if one feedback strategy had a larger impact on the gait stability of the users. The P-values for this comparison are in Table IV. For stride length, six subjects performed better with the CNST feedback and three subjects performed better with the CORR feedback. For step width, three subjects showed less variance with the CNST, three subjects showed less variance with CORR and three subjects showed no significant difference between the feedback strategies. For trunk sway, all nine subjects had less variance with the CORR feedback configuration. The trunk sway is visually compared in Fig. 2d. The two strategies appear to have the same standard deviation, suggesting that the two are equally effective at providing feedback to the user.

TABLE IV
P-VALUES OF LEFT-TAILED F-TEST FOR CORR VS CNST

Subject	SLV	SWV	TSV
01	1	1	$\ll 0.05$
02	$\ll 0.05$	$\ll 0.05$	$\ll 0.05$
03	$\ll 0.05$	0.88	$\ll 0.05$
04	0.61	$\ll 0.05$	$\ll 0.05$
05	0.21	$\ll 0.05$	$\ll 0.05$
06	$\ll 0.05$	1	$\ll 0.05$
07	1	0.82	$\ll 0.05$
08	$\ll 0.05$	$\ll 0.05$	$\ll 0.05$
09	0.29	$\ll 0.05$	$\ll 0.05$
Lumped	$\ll 0.05$	1	$\ll 0.05$

D. Cross Subject Analysis

To examine across subjects, a single lumped subject was created from all of the data (Tables I-IV). Each subject was normalized before being combined with the other subjects in order to compensate for the subject's position on the treadmill, such as standing very close to the edge vs standing near the treadmill control panel. This was done by subtracting the median value of the data for each subject to center each dataset about zero. The median value was subtracted instead of the mean value because the mean is more susceptible to outlying values (i.e. the data did not have a completely normal distribution) and the median better summarized the typical value for the different variables.

For the lumped subject, all measures indicated that NF was less stable than CTRL, verifying their sensitivity in this protocol. SLV and TSV improved under CORR, and SLV and SWV improved under CNST. The SWV P-value under CORR was close to 0.5 and the TSV P-value under CNST was close to 1, suggesting no significant improvement in those measures with the respective feedback strategy. These findings are similar to the majority of the individual subjects.

E. Questionnaire

In addition to the motion capture data collected, the subjects were given a questionnaire after the experiment to gauge their perception of the feedback device. The questions requested a numerical answer on a Likert scale (between 1 and 7). The subjects found both the CNST and CORR feedback strategies useful. All 9 subjects answered at or above neutral for questions regarding the helpfulness of the feedback strategies, CNST: 5.11 ± 0.93, CORR: 5.0 ± 1.50.

IV. Discussion and Conclusion

The CTRL trials were expected to have the least variability, the NF trials to have the most, and the CNST and CORR trials to fall within the two, i.e., var(CTRL) $<$ var(CNST) $\leq$ var(CORR) $<$ var(NF). As discussed previously, a decreased variability suggests improved stability and balance. If the feedback system improved the stability and balance of the user, it is expected that the variability would be lower with the feedback device than without. We first needed to verify that these measures are valid for assessing stability in our study. With the exception of two subjects, the CTRL displayed the least variability for SLV, SWV, and TSV. It is suspected that the two outlier subjects walked very cautiously on the prosthesis rather than trying to walk naturally. The subjects may have used the handrails or harness to support themselves, against the instructions given. This behavior may have deceived the stability measures, which we consider outlier cases.

For the majority of the subjects, the feedback strategies showed a decreased variability in step width and trunk sway compared to the NF trials. This data suggests that the feedback was assisting the subject in controlling the position of their prosthetic leg and torso in the medio-lateral direction (i.e. step width and trunk sway). The CORR feedback strategy improved more subjects than the CNST feedback strategy. The data showed that the CORR feedback strategy generally had better variability than the CNST feedback strategy in a direct comparison. Figs. 2b and 2c visually show that these conclusions appear to be correct.

The CNST feedback strategy is a direct mapping of the COP to the thigh. We hypothesized that this strategy would provide more phase-based information and thus enable superior gait stability than the CORR strategy. However, our results did not directly support this idea. Every subject displayed significantly better results with the CORR feedback than CNST feedback in TSV, the majority did significantly better with CORR feedback in respect to SLV, and about 1/3 of the subjects did significantly better with CORR with respect to SWV. Our initial interpretation of these results was that the subjects might have ignored or became desensitized to the feedback. Looking at the questionnaire results, there is no clear answer for this. On average, the subjects found both strategies about equally comfortable (CNST: 4.11 ± 1.4 vs CORR: 4.0 ± 1.2) and equally helpful (CNST: 5.11 ± 0.9 vs CORR: 5.0 ± 1.5). We cannot conclude that the subjects thought the vibrations were annoying or that they chose to ignore them. Regardless, both feedback strategies facilitated improvements over the NF case.

The SLV corresponds to the anterior-posterior (A-P) direction, whereas TSV and SWV correspond to the medial-lateral (M-L) direction. More subjects improved in SWV than SLV when using either haptic feedback strategy. Does this mean that feedback related to the COP primarily contributes towards stability in the M-L direction? The toe-off and heel-strike events act as indicators for the beginning and end of the step. It is possible that these event indicators are sufficient for prosthesis users to maintain stability in the A-P direction, in which case the extra phase-based feedback provided by the CNST strategy may be unnecessary. However, there are no event indicators in the M-L direction. Both feedback strategies provided assistance in the M-L direction, which enabled significant improvements in SWV and TSV.

Although we only reported results of SLV for the sound leg, we also measured SLV for the prosthetic leg. In this case we only found significant improvements in 2 subjects. This makes sense because the feedback was never activated while the prosthetic leg was swinging. The feedback is based on the COP of the prosthetic foot, which does not directly influence the placement of the prosthetic leg. Instead, the prosthesis COP feedback contributed to the placement of the sound leg.

Our results generally agree with findings in other examples in the literature. Lee found that visual-auditory proprioceptive biofeedback enables amputee subjects to use their prosthetic limb for longer periods of time walking [21]. Peterka found that vibrotactile feedback on the torso consistently and significantly reduced body sway levels in amputees [24]. We found that SLV was generally improved with the use of our haptic feedback, but SWV improved in more subjects. This aligns with Owings et al., who found that examination of step width variability was a better discriminator between healthy young and older adults than stride length or step time variabilities on a treadmill [34]. Moe-Nilssen et al. found that trunk variability better classified “fit” and “frail” older adults than SWV [35]. While those are different categories of walkers than our subjects, a similar comparison can be made between prosthesis users and able-bodied walkers. In our study, more subjects improved their SWV than TSV, but some did improve TSV. The disagreement with [35] could be associated with their use of an accelerometer rather than marker position data for measuring trunk variability.

Our study involved able-bodied subjects using a prosthetic leg rather than amputee subjects. It was assumed that the able-bodied users would perform like novice prosthetic leg users, having worse stability than amputees who have used a prosthetic leg for years. Arguably, novice users would have the most need for the proposed haptic device. Our results were mostly positive, suggesting that the feedback was useful and improved the stability of novice prosthetic leg users. Experiments should be conducted using expert prosthetic leg users (i.e., amputees who can ambulate independently) to determine whether the benefits generalize to that population.

This work introduced a new way to provide balance assistance for transfemoral amputees during gait. The diminished senses of proprioception and gait cycle phase during prosthetic stance were assumed to be causes of inferior gait stability in amputees. We developed a vibrotactile feedback device that provides phase-based proprioceptive feedback to the user based on the COP location of the prosthetic foot. The device was shown to reduce variability in stride length, step width, and trunk sway in novice prosthetic leg users. From these results we might infer that the COP is an important part of proprioception that should be restored in transfemoral amputees. Less significant changes in SLV may have been observed because heel-strike and toe-off events already provide information in the A-P direction. Prosthetic leg users lack any such indicators in the M-L direction. The obvious next step is to retrofit the device to work with amputees and to prove that the improvements translate well. Another potential testing group could be those with peripheral neuropathy. This condition, which is commonly associated with diabetes, results in the loss of sensation in the distal extremities like feet. Assuming the potential subjects still have feeling in the thigh, our feedback system could be used without modification. Our results should directly translate over to this group of people, however experiments would be necessary to confirm this.

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