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A Preliminary Study of the Effects of Active Recovery Reflexes on Stumble Recovery in a Swing-Assist Knee Prosthesis

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Abstract— **This paper explores the effects of a swing phase stumble recovery controller in a swing-assist prosthesis. The prosthesis detects a stumble event and employs either a lowering recovery response – wherein the user's swing is truncated, and the leg is prepared for loading– or an elevating recovery response – wherein an amplified swing flexion is employed to step over the obstacle causing the perturbation. The controller described in this paper choses which of these responses to use based on the perturbation timing within the gait cycle, where stumble events which occur prior to an estimated 35 percent of the way through swing trigger an elevating response, and later stumbles trigger a lowering response. The potential efficacy of this approach was assessed in a preliminary study with two participants with transfemoral amputation; wherein each participant's walking was perturbed in early, mid, and late swing phase when wearing both their prescribed prosthesis and the swing-assist prosthesis prototype. When wearing the swing-assist device, 0 of the 13 perturbations resulted in falls, with none of the trials being classifiable as "near falls". Conversely, when using their prescribed device, one participant had a fall rate of 3 out of 6 perturbations, with 1 of the 3 recoveries being classifiable as a "near fall"; the second participant had a fall rate of 0 of 3 trials, with 2 of the 3 recoveries being classifiable as "near falls". For both participants, when recovery was achieved, it was accompanied by significantly longer periods of irregularity and asymmetry in gait when using their prescribed devices, as compared to the test device. These results suggest the possibility of substantial benefit provided by a low-power, reflex-based stumble recovery feature in knee prostheses.**

I. Introduction

There are an estimated 300,000 persons in the United States with transfemoral amputation[1][2], with the global prevalence approximated to be between 20 and 30 times higher[3]. Persons with transfemoral amputation have been shown to be at significantly higher risk of falling in daily life than their age matched able-bodied counterparts, with more than 60% of transfemoral prosthesis users reporting at least one fall in the preceding year [4][5][6]. Such falls often lead to significant injury and related medical costs for the affected population[4][7][8]. In addition to the physical and financial burden such falls place on prosthesis users, a history of falls and/or a fear of falling has been shown to have a significant impact on prosthesis users' social and emotional wellbeing [4][6][9].

When gait is perturbed, there are three responses used by able-bodied persons. Early swing perturbations (< 35-40% swing) typically elicit an elevating response, wherein the stumbled person reflexively flexes their knee further after the perturbation to clear the obstacle and extends their stride in the process. Elevating recoveries are characterized by high angular velocity of the knee, where the knee torque is directly related to the inertia of the leg. Late swing perturbations (> 50-60% swing) are generally addressed with a lowering response, with the person extending the knee and loading the foot prior to clearing the obstacle. This shortens the stride, and the person is then forced to take a longer subsequent stride, often with increased knee flexion to ensure obstacle clearance. During the stumble stride, lowering responses require a high knee torque to support body weight in stance, and a relatively low angular velocity. Most able-bodied persons will exhibit a third response, delayed lowering, for some subset of perturbations in the liminal space between elevating and lowering (35%-50% swing). This response begins as an elevating response, but then switches to a lowering response, such that the hip and knee extend in order to lower and load the foot. This delayed lowering recovery is characterized by a high-speed, low torque profile during the elevating period and high torque, relatively low speed movement during the lowering phase. These three operations are the basis of able-bodied persons' robust stumble recovery abilities [10][11].

Knee prostheses can be broadly divided into three categories: passive knees, microprocessor-modulated passive knees (also called MPKs), and powered knees. Passive devices and MPKs make up the vast majority of prescribed knee prostheses; powered (or robotic) knees exist on the commercial market, but are currently more prevalent within the research community (e.g., [12]–[19]) Different devices within each of these three categories offer varying levels of robustness when perturbed. Broadly speaking, MPKs can offer adequate late-swing perturbation recovery, due to their ability to generate large resistive impedances, allowing the user to perform a lowering response. However, due to their inability to add power, they are generally unable to perform an elevating recovery without substantial compensatory actions from the user. Passive knees, on the other hand, are generally less capable of performing lowering recoveries reliably due to their inability to volitionally increase their impedance. Though they are not powered, some of these devices are low enough impedance that an elevating response can sometimes be achieved using the impulse from perturbation, albeit non-reliably. Powered knees can theoretically offer both elevating and lowering responses; meaning that with proper controllers and detection, they should enable their user to recover from perturbations at various points in the gait cycle more effectively. In this work, the authors aim to evaluate if the

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addition of low amounts of active power in the form of a stumble reflex response can benefit the recovery of prosthesis users.

II. Swing-Assist Prosthesis Prototype

The work described in this paper was performed using a Stance-Controlled, Swing-Assisted (SCSA) Knee prototype, shown in Fig.1. This prototype employs a pair of parallel actuation systems – a modulated passive hydraulic system capable of high resistive torque and power dissipation, and a low-power brushless-motor drive system, capable of relatively lower torque and power addition. The two systems are integrated into a single actuator and cooperatively provide both high torque controllable resistance (via modulation of a two-way servovalve), and low torque controllable active assistance (via current control of the brushless motor). This device, the design of which is described in more detail in [20], also includes a suite of sensors, including an IMU, a load cell, and absolute and incremental encoders for controlling the knee and sensing user's movement and intent. It is of note for control purposes that the hydraulic damping valve is in parallel with a check valve (see Fig. 1c), which ensures that the hydraulic system is free to move in extension, even if the servo valve is locked. Additionally, the motor is used at times to provide passive damping as described in [21] to fine tune resistance in both extension and flexion.

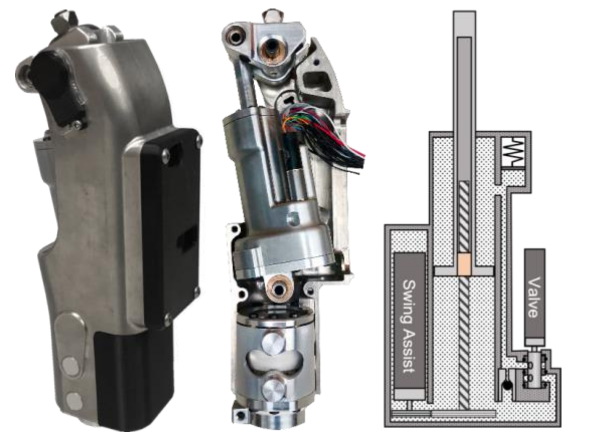

Figure 1: SCSA knee prototype. The leftmost image shows the device fully assembled, the middle with half of the housing removed, and the rightmost details the actuator diagrammatically.

III. Stumble Controller

A. Detection

A walking controller described in [22] is used to provide appropriate stance-phase stability and swing-phase movement when walking. Detection of a stumble event is limited to swing phase of the walking controller. Effectively, this means that stumble perturbation can only be detected when walking has been detected and the leg is unloaded. While these conditions are met, angular acceleration about the knee joint (via differentiation of the angular velocity signal from the gyroscope) and linear acceleration normal to the shank segment (via accelerometer) are monitored for perturbation detection. If a large angular acceleration in the knee flexion direction is detected at the same time a large rearward acceleration is detected, stumble is flagged and the estimated percent swing is recorded and used to select the recovery response (i.e., elevating or lowering).

B. Decision

When a stumble occurs, the determination of performing an elevating or lowering response was made by using the estimated swing percentage at which the perturbation occurred. Swing percentage was estimated staring at each toe off based on the previous swing time durations and the

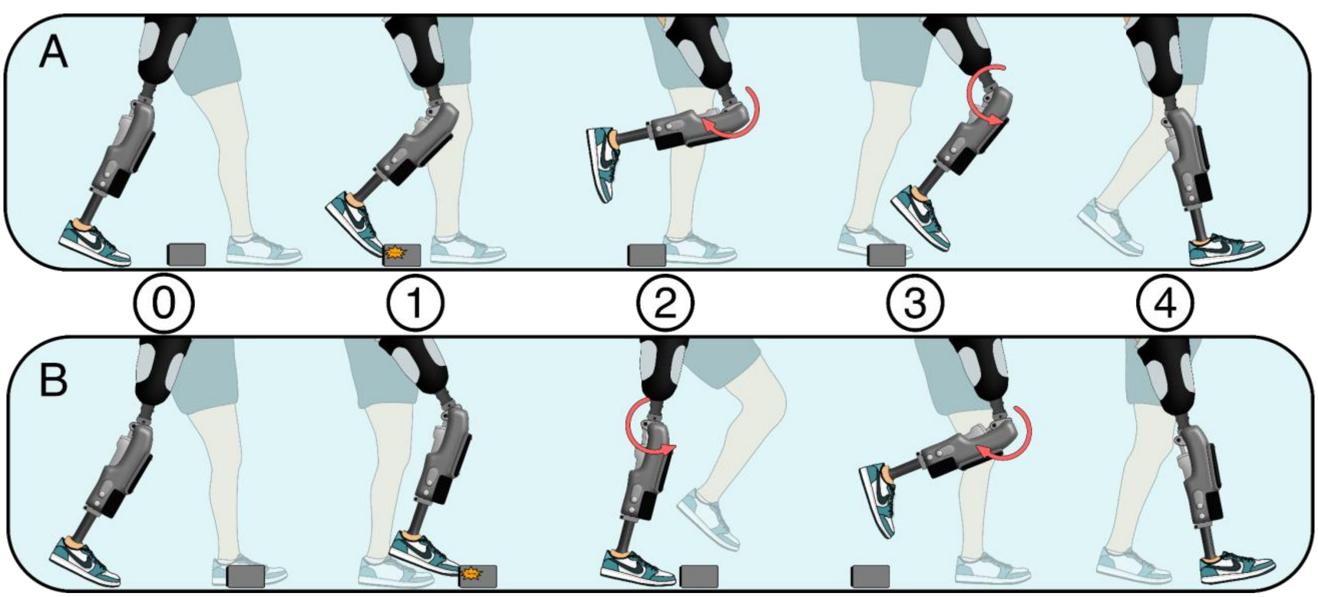

Figure 2: Bimodal stumble recovery responses: A) Illustration of an elevating response and B) a lowering response. For both responses, frame 0 shows walking pre-stumble, and frame 1 shows the incidence of stumble. In the case of the elevating recovery strategy, the leg first provides a torque to further flex the leg, clearing the obstacle (2A). Once the device is adequately flexed, it then extends (3A), and normal walking is resumed (4A). In the case of the lowering approach, when stumble is detected, the leg provides an extensive torque (2B) and locks the valve to allow for stance support. Then, on the subsequent stride (3B), it provides additional flexive torque to allow the user to easily clear the obstacle. It then resumes normal walking (4B).

preceding stance time duration. When a stumble occurs, the recovery approach is determined based on the estimate of percent swing at which the perturbation is detected. If the perturbation occurs prior to 35% of the way through swing, an elevating approach is adopted, as shown in figure 2A; otherwise, a lowering strategy, as illustrated in figure 2B is used. In the case of MPKs, a lowering response is employed throughout the whole of swing phase. As such, the decision approach employed here essentially alters the standard MPK response to replace the lowering response employed in the first 35% of swing phase with an elevating response instead.

C. Elevating Response

In an early stumble event, the SCSA hydraulic valve is already in a relatively open (i.e., low resistance) state at the time of perturbation (as per the walking controller). Regardless, at the detection of early stumble, the valve is commanded to an open (i.e., low resistance) configuration to ensure minimal resistance to knee flexion. In this case, the low impedance of the device is useful for two reasons: firstly, it allows the knee to use the natural dynamics of the collision to initiate knee flexion, and secondly, it reduces the angular momentum imparted to the user's center of mass as a result of the impact. This latter benefit helps to decrease the pitching forward of the user's trunk, facilitating recovery.

In addition to the opening of the valve, at perturbation detection, the drive system is commanded to provide one period of a square wave feed forward current with amplitudes of ± 8.5 Amps (this corresponds to a maximum torque of approximately 7.5 Nm at the knee but is non-constant because the slider crank transmission ratio is a function of knee angle). The torque pulse is timed to the estimated leg movement required to clear the obstacle based on cadence, wherein the crossover from flexion torque to extension torque occurs at 30% of the way through the pulse.

When the torque pulse switches from flexion to extension, the hydraulic valve (controlling resistance to flexion) is rotated to a locked configuration in preparation for heel strike. As previously mentioned, due to the presence of a check valve in parallel with the servo valve, this change does not affect the knee's resistance to extension.

As the knee completes its extension (below 15 deg flexion with an extensive velocity), the torque pulse is disabled, and the motor is instead used as a passive damper to mitigate terminal impact of the knee against the hard stop. Upon heel strike, normal walking control is resumed.

D. Lowering Response

Upon detecting a late swing stumble perturbation, the SCSA first commands the hydraulic valve to a fully locked position. Again, due to the presence of the parallel check valve, the knee is still able to freely extend, but will now be locked from flexing further. At the same time, a stiff, virtual spring is emulated at the knee with its equilibrium angle set to full extension. This assists the user in quickly moving the leg into as stable a position as possible. The output of this virtual spring is software-limited to 10 Amps, approximately 8.75 Nm about the knee, at peak transmission.

This is effectively the approach that an MPK performs to a stumble perturbation, except with greater extension assistance. A key difference between the SCSA approach to lowering versus an MPK, however, is in the following step, wherein the SCSA walking controller is modified to increase knee flexion above the values of normal walking, allowing the user to clear the stumble obstacle more easily without requiring any compensatory movements.

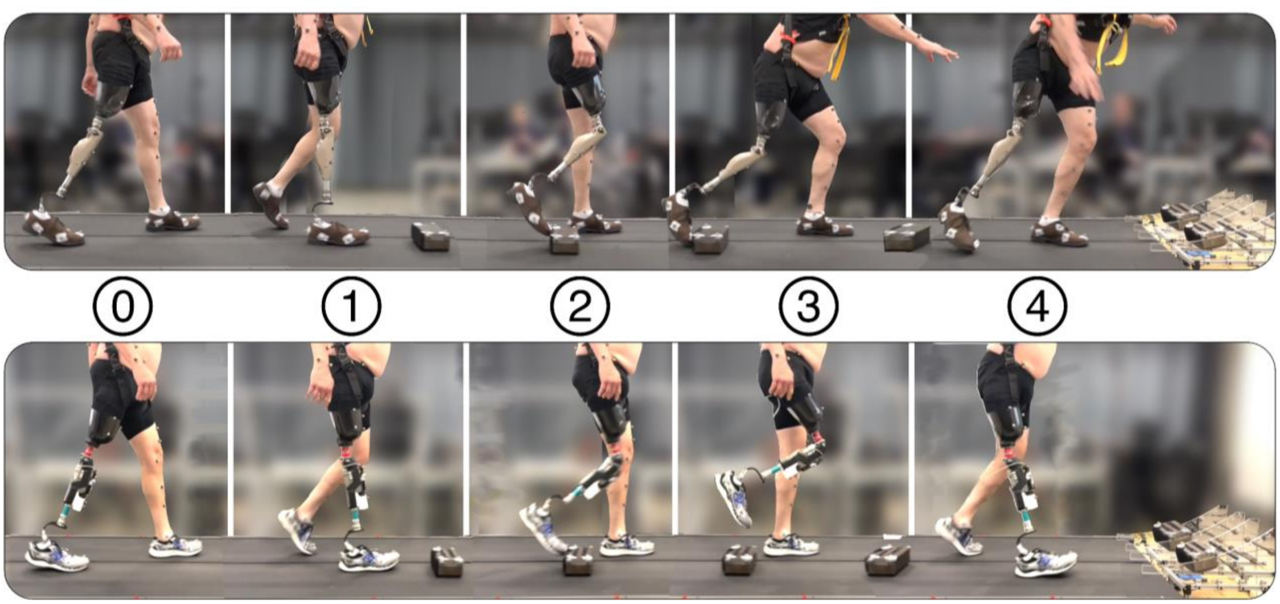

Figure 3: Experimental set up and example stumble. P2 shown being stumbled in early swing on his prescribed C-Leg 4 (A) and on the SCSA Knee (B). For both image sets, the incidence of stumble occurs at frame 2. On his prescribed device, the participant's leg does not clear the obstacle (3A), and he must rotate his body (4A) to get past the block. Subsequently, the participant hopped sideways for several seconds until regaining normal walking gait after 6 strides (Note Table 2).This was classified as a near fall due to his increased sagittal and transverse plane torso angles. In the lower image set, the SCSA's flexive torque pulse allows the user to clear the obstacle (3B) and immediately resume normal walking (4B).

IV. Experimental Assessment

To assess the prospective efficacy of these recovery reflexes, a two-participant study was performed. In it, the participants wore 80 markers which allowed for tracking of their upper and lower body movements via motion capture (Vicon, Oxford, GBR). The participants then walked at 0.8 meters per second on a split belt, force-instrumented treadmill (Bertec, Columbus, USA) while wearing passive noise-canceling headphones overtop earbuds playing white noise, inferior visual field obstruction glasses, and a force-instrumented safety harness, as described in [23]. At the same time, the participants counted backwards by intervals of seven from a provided number to ensure that they were mentally distracted and would not fixate on their gait. After a randomized, predetermined number of strides had occurred to guarantee steady-state walking, the perturbation system developed in [23] was used to induce a stumble on the participant's prosthetic limb. If the harness was loaded with a force equivalent to 50% of the participant's bodyweight, the trial was marked as a fall.

For both participants, this experiment was performed first on their prescribed devices (Blatchford KX06 for P1, and C-Leg 4 for P2), and subsequently was performed on the Vanderbilt SCSA. An uneven number of trials was performed for each device, since the stumble trials were taxing for the participants, and data collection was stopped if they felt too worn out or unstable on their device. On a day prior to the presented stumble test on the SCSA, each participant was allowed to walk on the device and performed several trials of stumbling on the device while donning various levels of the sensory occlusion gear, to their comfort. After the trials were performed, motion capture information was processed using Visual 3D (C-Motion, Germantown, MD) to generate kinematics data for the trials. The experimental setup and an example trial is shown in Fig. 3, with the perturbation setup shown in the far right of each series.

Table I: Stumble Outcome Metrics

	Percent Swing (Post Processing Estimation)						
P1 Prescribed	22	23	25	39	56	68	
P1 SCSA	21	28	41	42	55	70	
P2 Prescribed	29	41	62				
P2 SCSA	22	23	27	62	63	64	67

V. Results & Discussion

This study comprised 9 stumbles on prescribed devices (6 for P1, 3 for P2) and 13 stumbles on the reflex enabled SCSA device (6 for P1, 7 for P2). The discrepancy in number of trials, as well as the low overall trial count is a result of mental and physical exhaustion on the part of the participants. It is of note that near falls and complete falls were especially taxing, which allowed for fewer stumble trials on the daily use devices. The recovery outcome results of these trials are shown in Table I. Each cell's contents indicate the percent swing at which the perturbation occurred, while its color indicates the result of the perturbation. Red cells indicate a perturbation resulting in the participant falling, as measured by the instrumented safety harness (i.e., greater than 50% body weight); yellow cells indicate that the participant recovered but achieved a sagittal plane torso angle or a transverse plane torso angle greater than 60 deg and was thus classified as a "near fall"; and green cells indicate that the participant successfully recovered with a torso angle less than 60 deg. This same data is presented more visually in Fig. 4. It is worth noting that any perturbation prior to 40% swing is considered an "Early Stumble" for the purposes of this paper, while "Late Stumbles" comprise perturbations after 60%, and the remaining 20% are classified as "Middle Stumbles". Recall that the SCSA knee employed a lowering response for all middle and late perturbations, and an elevating response for all early perturbations.

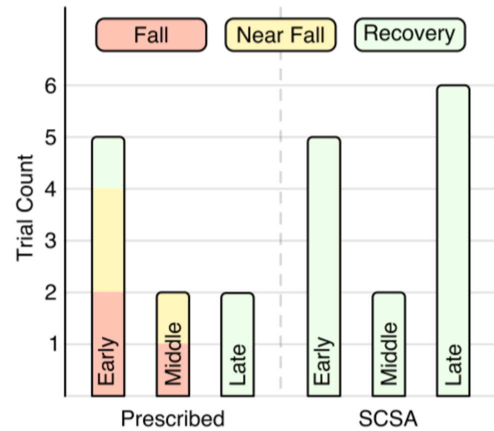

Figure 4: Stumble outcomes: results of all stumble trials summarized with falls being indicated in red (3/9 on prescribed, 0/13 on SCSA), near falls indicated in yellow (3/9 on prescribe, 0/13 on SCSA), and successful recoveries indicated in green (3/9 on prescribed, 13/13 on SCSA)

For the 6 prescribed trials and 13 SCSA trials in which the participants recovered from the perturbation, analysis of metrics associated with quality of recovery were additionally analyzed. An overview of the peak torso rotation both in the sagittal plane (i.e., forward lean) and in the transverse plane (i.e., twisting of the body) is shown for all trials in Fig. 5. As indicated in the figure, use of reflexes decreased these indicators of fall likelihood, and reduced them to values exhibited by able-bodied individuals [22].

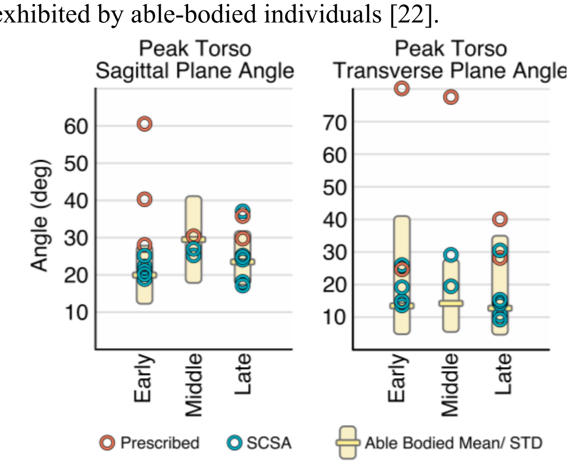

Figure 5: Recovery outcome torso metrics: Data from recoveries on prescribed device shown in red, recoveries on the SCSA device in blue, and able-bodied mean and standard deviation data shown in yellow

In addition to improved metrics of trunk stability, regular hip and knee walking patterns were regained by the users notably sooner after perturbation when using active reflexes, as compared to their passive, prescribed knees. Representative knee angle trajectories following a stumble for both early and late swing perturbations are shown for Participant 1 and 2 in figures 6 and 7, respectively. Additionally, the number of strides between the stumble perturbation and achieving 80% of typical peak knee flexion are shown in Table II. In all trials, the SCSA reached a peak knee angle at or above 80% of its prior steps (average peak knee angle from the preceding 10 swing phases) in the swing phase immediately following the perturbation. However, on the prescribed devices it took an average of nearly 4 strides before unperturbed knee trajectories were attained.

In a non-laboratory setting, this prolonged diminished knee trajectory would presumably make the user more susceptible to subsequent scuffs and/or stumbles, especially on uneven terrain. This is especially detrimental in late swing stumbles, where the subsequent stride requires *greater* knee flexion than normal to clear the obstacle without requiring compensatory behaviors.

Table II: Recovery Outcome Metrics

	Strides Post Stumble to regain 80% Peak Knee Angle						
P1 Prescribed	NA	NA	2	2	NA	1	
P1 SCSA	1	1	1	1	1	1	
P2 Prescribed	6	6	> 5*				
P2 SCSA	1	1	1	1	1	1	1

*indicates that the treadmill was decelerated before participant achieved 80% knee flexion

VI. Conclusion

This paper describes the implementation and assessment of stumble reflex recovery responses in a prototype knee prosthesis. Specifically, a two-participant study indicated that relatively small amounts of power added following stumble perturbations decreased the number of falls experienced by the users, and improved substantially the quality of recovery and the amount of time it took to return to non-perturbed walking.

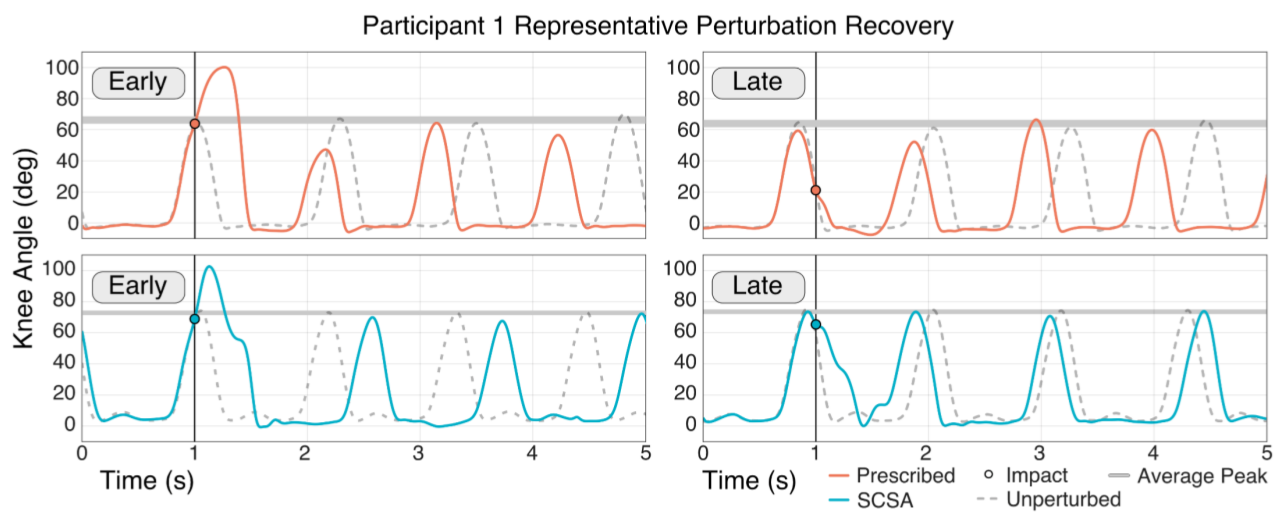

Figure 6: Representative recoveries for early and late stumbles for P1: Prescribed recoveries for early (left) and late (right) shown for respective devices, with prescribed trials above in red and SCSA trials shown below in blue. Representative non-perturbed strides are shown in a grey dashed line, the time of impact shown with a colored bubble and a vertical black line. Grey horizontal bands show mean peak knee angle plus and minus one standard deviation.

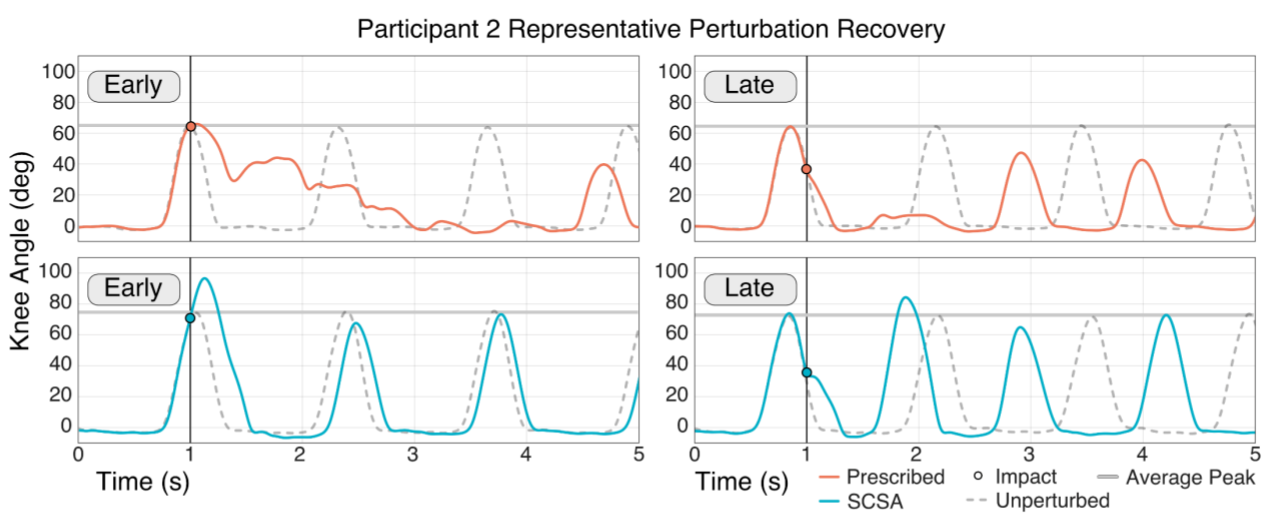

Figure 7: Representative recoveries for early and late stumbles for P2

Though this was a limited study, it points to the value of future studies assessing the prospective value of stumble recovery reflexes in semi-powered and fully powered knee prostheses. Further, while this study demonstrated the benefits of a simple swing-percentage-based response decision, other more sophisticated decision algorithms might enable further improvements in recovery robustness.

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