Development and Application of a Virtual Reality Stumble Method to Test an Angular Velocity Control Orthosis

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Abstract

Stance control orthoses stabilize and support users during weight bearing while permitting free knee motion during swing. An angular-velocity control knee joint (Ottawalk-Speed, OWS), prevents knee collapse during a stumble while offering a lightweight and compact design. The purpose of this study was to develop a virtual perturbation to measure OWS response to a stumble, when worn by knee-ankle foot orthosis (KAFO) users.

A split speed perturbation, developed for the CAREN-Extended system, was tested on five able-bodied participants during treadmill level walking. Three KAFO users, with the OWS, were subjected to the perturbation during five level walking trials. Kinematic data were collected at 120 Hz during testing and OWS user feedback was collected via a questionnaire. Lower body joint kinematics and questionnaire responses were evaluated.

Two recovery strategies were observed for able-bodied participants: hopping and stopping. Knee flexion increased at recovery weight acceptance for all trials. OWS participants used a straight-legged strategy to recover from the split speed perturbation, with an extended knee throughout weight acceptance. OWS questionnaire evaluation revealed positive responses to walking ability, with ease of use receiving some negative comments.

The split speed perturbation was not able to produce sufficient knee flexion upon landing for the KAFO users, for OWS stumble evaluation. The OWS was successful in allowing free knee motion when walking on level, incline, and uneven surfaces. Further study is required to measure OWS response in stumble situations typically experienced by the target population.

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List of Abbreviations

CAREN: Computer Assisted Rehabilitation Environment

DOF: Degrees of Freedom

DST: Double support time

FO: Foot off

FS: Foot strike

KAFO: Knee Ankle Foot Orthosis

OWS: Ottawalk-Speed

P&O: Prosthetics & Orthotics

PERTFS: Perturbation step foot strike

PERTFO: Perturbation step foot off

REC1FO: Recovery 1 step foot off

REC1FS: Recovery 1 step foot strike

REC2FO: Recovery 2 step foot off

REC2FS: Recovery 2 step foot strike

RVR: Rehabilitation Virtual Reality

SCKAFO: Stance Control Knee Ankle Foot Orthosis

TOHRC: The Ottawa Hospital Rehabilitation Centre

List of Definitions

Braced leg: For OWS participants, this refers to the leg with quadriceps weakness for which the orthosis was prescribed. Though quadriceps weakness may be present in both legs, the affected or braced leg refers to the leg with the orthosis.

CAREN-Extended (Computer Assisted Rehabilitation Environment): A virtual reality environment consisting of a motion platform, dual track instrumented treadmill, projection screen, and Vicon motion capture system

Digitizing wand: A tool used to identify anatomical landmarks for motion analysis. This wand contains a spring tip and four reflective markers at specific orientations. When the wand tip is depressed, a digitized marker is recognized in Visual 3D.

Digitizing trial: A short motion capture trial used to identify anatomical landmarks. The digitizing trial is used in conjunction with a static trial during post processing to orient limbs within the motion capture area and define link segments for analysis.

Digitized markers: Markers representing anatomical landmarks that are defined using a digitizing wand during digitizing trials. Digital points are useful for reducing the number of reflective markers placed on participants during motion analysis, which can reduce instances of lost or dropped markers.

Double support time (DST): Period in the gait cycle with both feet on the ground. Occurs at the beginning and end of stance phase

Donning/doffing: The act of putting on or taking off an orthosis.

Fall: If a participant could not recover from a stumble situation, a fall occurred if the CAREN safety harness was pulled taught, indicating body weight support by the harness.

Foot flat: Following foot strike, the period at the start of the stance phase when the plantar foot makes full contact with the ground.

Foot strike: Event in the gait cycle when foot contact is made with the ground. This marks the transition from swing phase to stance phase.

Foot off: Event in the gait when the foot leaves the ground. This marks the transition from stance phase to swing phase.

Free knee motion: Referring to the braced leg of OWS participants, when knee joint motion is not restricted by the OWS during gait. If the joint does not engage during gait, free knee motion is achieved.

Gait cycle: Sequence of events between two sequential foot strikes by the same limb, consisting of a stance phase and swing phase.

Knee ankle foot orthosis (KAFO): A lower limb brace that extends from the foot to the thigh, with ankle and knee joints.

Knee collapse: A sudden giving out or buckling of the knee.

Kinematics: The branch of mechanics concerned with the motion of objects without reference to the forces that cause the motion.

Treadmill level (TL) gait: For each subject and each trial, kinematic data were extracted from three treadmill level walking strides prior to the perturbation. Values for each participant were averaged across trials to represent each participant's typical TL gait values.

Non-braced leg: For OWS participants, this refers to the leg without an orthosis. The quadriceps may still possess some deficits but are stronger than those of the affected leg.

Ottawalk Speed (OWS): An angular velocity control approach for stance control orthoses that engages to resist knee flexion when knee flexion angular velocity surpasses a preset threshold. This joint uses hydraulic technology and is based on the premise that knee flexion velocity in an unexpected stumble situation surpasses knee flexion velocity during gait.

Split-speed perturbation: The perturbation method developed within the CAREN-Extended system during this project and used for OWS testing. In self-paced treadmill mode, the perturbation was applied at mid-swing and caused one treadmill belt to speed up while the other slowed down, before returning to a linked treadmill speed.

Stance control knee ankle foot orthosis (SCKAFO): A KAFO prescribed to people with lower limb muscle strength deficits that allows free knee motion during swing. These orthoses are typically prescribed to people with hip strength grades of 3 or greater.

Stance phase: The period in the gait cycle when a foot is in contact with the floor.

Static trial: A short motion capture trial with the person standing still in a known posture that is used to identify tracking marker locations. A static trial could also include anatomical landmarks or these landmarks could be obtained from a Digitizing trial

Stride: The interval in the gait cycle between two sequential initial contacts of the same foot (i.e. right to right).

Stumble: To miss one's step or momentarily lose one's balance; trip and almost fall. In this thesis, a stumble purposely induced in the CAREN environment was considered successful if knee flexion of the predetermined leg increased during weight acceptance, compared to TL.

Stumble Recovery: The period following a stumble perturbation, during which the participant regains balance before resuming their typical gait pattern. Kinematic data following perturbation were described for the perturbation stride, recovery 1 stride, and recovery 2 step.

Swing: The period in the gait cycle when the foot is not in contact with the floor

Weight acceptance: The initial gait cycle period when body weight is loaded onto the limb.

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Chapter 1. Introduction

Lower limb weakness, particularly knee extensor weakness, is a debilitating condition that alters normal gait patterns and contributes to an increased risk of knee collapse. (Bohannon, 1986; Lohmann Siegel, Stanhope & Caldwell, 1993; Lohmann Siegel, Kepple & Stanhope, 2006). Over time, this condition may result in reduced feelings of security and comfort during daily tasks, activity avoidance, and a diminished sense of well-being (Lucas, 2007; Peethambaran, 2000).

Knee ankle foot orthoses (KAFO) are prescribed to populations with substantial quadriceps weakness (Irby, Bernhardt & Kaufman, 2007; Irby, Bernhardt & Kaufman, 2005; Kaufman, Irby, Basford & Sutherland, 2001). Locked-knee KAFOs cross the ankle and knee joints to lock the affected knee in extension, thereby improving user security, stability, and function (Herbert, 2006). However, these devices force users to ambulate with stiff-knee gait, leading them to develop undesirable compensatory strategies that may result in chronic pain and loss of motion over time (McMillan, Kendrick, Michael, Aronson & Horton, 2004; Sulzer, Gordon, Dhaher, Peshkin & Patton, 2010).

Rehabilitation researchers and engineers have worked to improve conventional KAFOs by developing stance control orthoses (SCO). SCO designs allow free knee motion during the swing phase of gait and while preventing limb collapse during weight-bearing (Lemaire, Goudreau, & Yakimovich, 2009). Stance control devices integrated with KAFO designs are known as stance control knee ankle foot orthoses (SCKAFOs) and have been shown to reduce compensatory strategies associated with KAFO use (Herbert & Liggins, 2005; Irby, Bernhardt, & Kaufman, 2005; McMillan et al., 2004). Unfortunately, current commercial SCKAFOs are often noisy, bulky, heavy, expensive, and in some cases, are not effective for fall prevention during unpredictable situations, such as a stumble.

To address these issues, researchers and engineers at TOHRC developed a new angular velocity control approach for stance control, named the Ottawalk-Speed (OWS) (Lemaire et al., 2009; Yakimovich, Lemaire & Kofman, 2006). This novel design used hydraulic technology to resist knee flexion when the knee flexion angular velocity surpassed a preset threshold. Unlike many commercial SCKAFO designs, the OWS is lightweight and compact, with potential to engage and prevent a knee collapse in unpredictable stumble situations. However, users cannot take advantage of the new technology until the device is tested on the target population.

A perturbation method was developed within the CAREN-Extended system (Computer Assisted Rehabilitation Environment). CAREN-Extended is a virtual reality (VR) environment that consists of a 180 degree projection screen, motion platform, instrumented dual-track treadmill, and motion capture system. This thesis details the development and evaluation of a CAREN-based stumble perturbation method and use of this method for testing the OWS orthosis with a KAFO user population.

1.1. Rationale

1.1.1 Stumbling

Research has shown that repeated trip or slip training can improve recovery strategies in populations at increased risk of falls, and that these improvements can be retained for several months (Pai & Bhatt, 2007; MacCulloch, Gardner, Bonner, 2007; Parijat, Lockhart, 2011). A stumble perturbation that causes increased knee flexion during recovery is beneficial when incorporated into clinical rehabilitation practices, or for future projects related to fall and balance research.

The CAREN-Extended system is ideal for the development of a stumble method. This virtual environment has the capacity to create repeatable and controllable methods of perturbation within a safe environment. The VR stumble protocol created for this thesis is used regularly for physical rehabilitation training at TOHRC. Adjustable perturbation settings make it possible to accommodate different populations who may be exposed to this stumble method.

1.1.2 OWS testing

The OWS allows free knee motion during gait, and provides security by resisting knee flexion during a knee collapse event. The lightweight and compact hydraulic design has the potential to improve upon typically bulky, heavy and noisy SCKAFOs currently on the market (Lemaire et al., 2009). The OWS also has the unique potential to be incorporated within a knee orthosis, benefiting people with limited or temporary knee extensor weakness that do not require a full SCKAFO.

The OWS has shown promise in initial reports involving able-bodied subjects, allowing free knee motion during level ground gait and engaging successfully to resist knee flexion in knee collapse tests. However, these results should not be generalized to the target clinical population, since orthosis users may not respond to the OWS in the same way as able-bodied participants. Additionally, it is not known how the OWS will react to a real world stumble, trip, slip, when worn by a person with lower extremity weakness. The angular velocity control approach requires knee flexion velocity to surpass a preset threshold in order to engage and resist knee flexion, for fall prevention, and it is uncertain whether this threshold will be reached following a stumble. Finally, the OWS has only been evaluated in level ground walking conditions; it is not known if the OWS joint will inappropriately engage when walking in different surface conditions, such as incline, decline, rolling hills and uneven ground. The OWS SCKAFO must be evaluated in these areas on a sample of orthosis users before being deemed suitable for commercial release.

1.2 Objectives

1. Develop a stumble method using the CAREN-Extended system that induces increased knee flexion during weight acceptance in able-bodied participants.

2. Determine if the OWS successfully engages to resist knee flexion following a stumble when worn by KAFO users.

3. Determine if the OWS allows free knee motion during gait when worn by KAFO users.

4. Understand orthosis user opinions of the OWS orthosis.

1.3 Hypotheses

1. A stumble that induces increased knee flexion during weight acceptance in able-bodied participants will cause a stumble with increased knee flexion during weight acceptance for OWS orthosis users with knee extensor weakness.

2. In the event of a knee collapse following the stumble perturbation or during ambulation on a variety of simulated surfaces (treadmill level, incline, rolling hills, uneven), the OWS joint will engage and resist knee flexion to prevent a fall.

3. The OWS joint will not restrict knee flexion or extension during gait.

1.4 Limitations

The population of orthosis users who fit the requirements for OWS use is limited. Limited recruitment possibilities resulted in a small sample size for OWS testing, which limited the result generalizability. This also affected stumble perturbation development since the authors did not want to expose OWS testing participants to the perturbation prior to the testing session. Able-bodied participants were recruited for stumble method development. Ideally, the same population of potential OWS users would have participated in stumble method development, since different recovery responses occurred between populations.

For OWS participants, training time with the OWS SCKAFO was limited. It is unlikely that participants fully accommodated to the device and questionnaire results related to user satisfaction may not reflect results that would be seen after long term OWS use.

Limited accommodation time to the CAREN environment could also be considered a limitation. RVR session time is both expensive and limited, and people may adapt to the CAREN environment at different rates. More accommodation time could result in a greater sense of ease and unreserved gait, especially when participants are aware that a stumble will randomly occur. This could affect the way participants recover from perturbations within CAREN, as compared to during real world stumble situations.

1.5 Overview of the Thesis

Chapter 2 reviews the literature, and outlines temporal spatial parameters and lower body joint kinematics on level ground surfaces for able-bodied, KAFO, and SCKAFO user gait; provides a design overview of current SCKAFOs and the OWS SCKAFO; and summarizes the literature for perturbation methods that induce a stumble. Chapter 3 describes the study design, including an overview of instrumentation, participants, experimental protocols, and data analysis. Results are reported in Chapter 4 and discussed in Chapter 5. Chapter 6 outlines conclusions drawn from study outcomes.

Chapter 2. Literature Review

2.1 Able-Bodied Walking Gait

The level ground gait patterns of able-bodied populations are well established in the literature. This has allowed researchers to establish a basis for human gait that can be compared with pathological gait. The ability to distinguish differences between able-bodied and pathological gait advances the understanding of conditions affecting locomotion and can help clinical rehabilitation programs improve function in daily living. Throughout the literature, researchers are in agreement concerning the phases that comprise one complete walking gait cycle (Gamble & Rose, 2006; Kerrigan, 1998; Perry, 1992; Whittle, 2002). Figure 1 illustrates gait phases, periods, and events during this cycle, and Table 1 describes these periods and their function. The gait cycle is divided into two phases: stance phase and swing phase. Stance phase occurs from foot strike (FS) to ipsilateral foot off (FO), and normally constitutes 0-62% of the gait cycle. Swing phase occurs from foot-off to ipsilateral

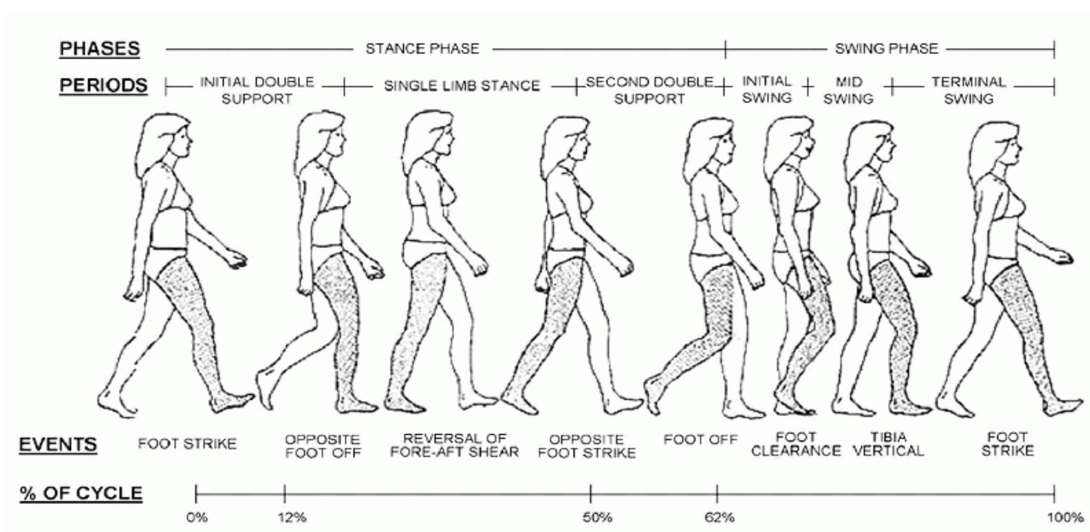


Figure 1: Gait cycle (Gamble & Rose, 2006).

foot strike and normally comprises the final 62-100% of the gait cycle. In the transition from stance to swing, there is a period of double support where both feet are in contact with the ground. Double support occurs in the first and last 10-12% of each stance phase, totalling 20-24% of a complete gait cycle (Gamble & Rose, 2006; Perry, 1992).

Table 1: Gait Cycle: periods and functions (Gamble & Rose, 2006).

Period	% Cycle	Function	Contralateral Limb
Initial double limb support	0-12	Loading, weight transfer	Unloading and preparing for swing (preswing)
Single limb support	12-50	Support of entire body weight; centre of mass moving forward	Swing
Second double limb support	50-62	Unloading and preparing for swing (preswing)	Loading, weight transfer
Initial swing	62-75	Foot clearance	Single limb support
Mid swing	75-85	Limb advances in front of body	Single limb support
Terminal swing	85-100	Limb deceleration, preparation for weight transfer	Single limb support

2.1.1 Temporal spatial parameters

Temporal-spatial gait parameters can tell us a lot about the timing, speed, and distance covered during walking. These parameters include cadence, step time, step length, stride length, and stride velocity. Stance or swing phase can also be represented as a percentage of the gait cycle. Table 2 displays average values of normal adult gait, as described by Perry (1992). Whittle (2002) reported similar findings for an age range of 18-64 years. Ranges for cadence were 82-135 steps/min in men and 97-138 steps/min in women, stride length were 1.22-1.85m for men and 1.04-1.58m for women, and walking velocity were 0.96-1.82m/s for men and 0.91-1.66m/s for women. Similar values for normal gait velocity can be found in reports by Bohannon (1996) and Oberg, Karsznia, & Oberg (1993). Sex can affect the temporal spatial gait parameters due to differences in height, weight, and age (Winter, 1991).

Table 2: Temporal spatial parameters of normal gait (Perry, 1992).

Temporal gait parameter	Value
Velocity (m/s)	1.37
Cadence (steps/min)	113
Stride Length (m)	1.41
Stance (percent of gait cycle)	62
Swing (percent of gait cycle)	38
Single Limb Support (percent per leg per gait cycle)	40
Double Support (percent per leg per gait cycle)	10

2.1.2 Kinematics

Kinematic measures describe the movement of persons or objects without reference to the forces responsible for the movement (Winter, 1991). Relative to the study of gait, kinematic measurements allow us to track joint and limb segment motion throughout the gait cycle. Sagittal plane motion curves for the ankle, knee, and hip throughout the normal gait cycle are shown in Figure 2. Similar curves have been presented by Kirtley (2006) and Winter & Robertson (1978). These graphs show that at foot strike, the ankle, knee and hip joints work together to absorb the weight of the falling body and transfer it forward through stance phase (Gamble & Rose, 2006; Perry, 1992). Following foot strike, during weight acceptance, the ipsilateral ankle plantar-flexes rapidly in the first 6% of the gait cycle, and knee flexion increases to approximately 21° in the first 12% of the gait cycle (Winter, 1991; Wittle, 2002). As the limb is loaded, the hip begins to abduct and the pelvis tilts upward on the ipsilateral side.

As stance phase progresses, the ankle dorsiflexes and the knee and hip extend to advance the body forward. At toe-off (60%) the ankle is rapidly plantar-flexed to -20° while the knee rapidly flexes to 64° (Perry, 1992). The hip flexes to 20° and adducts while the pelvis tilts downward and internally rotates on the ipsilateral side (Winter, 1991). As the limb progresses through swing phase (62-100%), the ankle gradually dorsiflexes to 9° and knee flexion slightly decreases and levels off. The hip extends to -10° of flexion and continues to adduct while the pelvis begins to drop and externally rotate (Gamble & Rose, 2006; Winter, 1991).

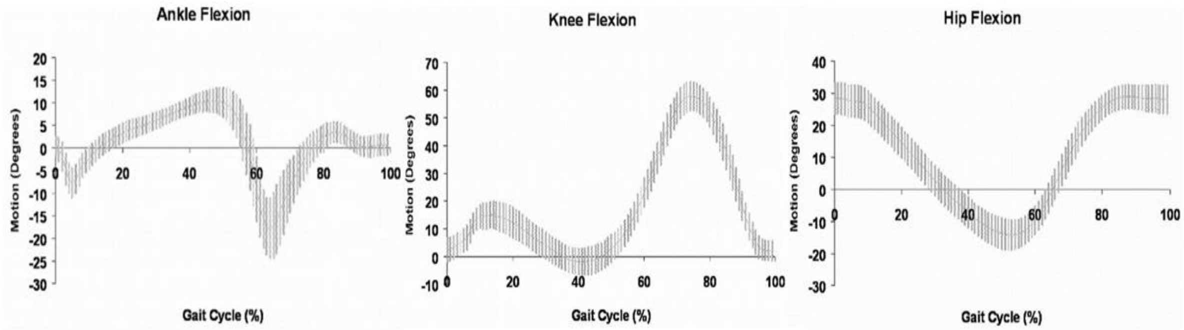


Figure 2: Ankle, knee, hip flexion during the gait cycle (Gamble & Rose, 2006).

2.2 Knee-Ankle-Foot-Orthoses

Studies have shown that populations with quadriceps weakness displayed gait adaptations 5-15 months post injury, as compensation for lower limb muscle strength deficits (Lohmann Siegel et al., 1993). Knee-ankle-foot orthoses (KAFO) are lower extremity bracing devices that extend over the knee, ankle, and foot. KAFOs prescribed to people with lower limb muscle weakness, particularly in the quadriceps (Bernhardt, Irby & Kaufman, 2006; Kaufman & Irby, 2006; Kaufman, Irby, Basford & Sutherland, 2001; McMillan et al., 2004). Numerous injuries, conditions, and disorders may cause muscle weakness; including, polio, stroke, incomplete spinal injury, neurovascular pathology and trauma, and neurological and developmental defects (Herbert & Liggins, 2005; Peethambaran, 2000; Irby et al.; Yakimoich et al., 2006; Yakimovich, Lemaire & Kofman, 2009).

The conventional KAFO (Figure 3) compensates for muscles that are too weak to function normally during daily activities and is typically used in cases when joint stability is more important than ease of mobility (Suga, Kameyama, Ogawa, Matsuura & Oka, 1998). These devices keep the lower limb locked in extension and forces users to walk with stiff-knee gait, thereby eliminating the need for knee extensors to bring the leg back into extension during swing phase and preventing knee movement into unrecoverable flexion during stance (Lohmann Siegel et al., 2006; Perry, 1992). However, stiff-knee gait causes users to adopt compensatory strategies that employ other muscle groups and, over time, may cause conditions of soft tissue and joint dysfunction, chronic pain, and loss of motion that affects the hips and lower back (McMillan et al., 2004; Yakimoich et al., 2006). These consequences, as well as factors related to comfort, reliability, ease of use, and cosmetic appeal, may cause users to abandon their device and sacrifice the opportunity for improvements to stability and mobility (Phillips & Zhao, 1993).

Figure 3: A conventional KAFO (Treasure State Orthotic and Prosthetic Clinic, 2011).

2.2.1 Temporal spatial parameters

KAFO user temporal spatial gait parameters differ from normal. Research by Irby et al. (Irby et al, 2005) measured velocity, cadence, stride length, and single limb support time for 21 KAFO users, of which 13 were experienced users (average 28 ± 18 years). Two years later Irby et al. (2007) measured the same parameters for 14 less experienced KAFO users.

Table 3 shows the average values reported by these two studies as well as the normal values reported by Perry (1992). All temporal spatial parameters for KAFO users were lower than able-bodied participants. Interestingly, the lowest values were found in the 2005 study, which reported a more experienced group of KAFO users than the 2007 study. Additionally, a report by Suga et al. (1998) showed that the stance phase was significantly reduced when healthy subjects wore a locked KAFO, compared to normal walking.

Table 3: Average temporal spatial parameters of KAFO and normal gait (Irby et al., 2007; Irby et al, 2005; Perry, 1992).

Temporal Spatial Parameter	KAFO (2005)	KAFO (2007)	Normal
Velocity (m/s)	0.553	0.72	1.36
Cadence (steps/m)	77	78	113
Stride Length (m)	0.86	1.07	1.41
Single limb support (% per leg per cycle)	27	31	40

2.2.2 Kinematics

Conventional KAFO users typically adjust to locked-knee gait by developing compensatory strategies in both braced and non-braced limbs, leading increased energy expenditure during daily activities (Bernhardt et al., 2006; Herbert, 2006; Irby, Kaufman, Wirta & Sutherland, 1999; Suga et al., 1998; Waters, Campbell, Thomas, Hugos & Davis, 1982; Zissimopoulos, Fatone & Gard, 2007). Irby et al. (2005) stated that the three most common adaptations related to straight-leg gait are hip hike, circumduction, and vaulting. Hip hiking is identified by pelvic obliquity during swing, specifically ipsilateral elevation of the iliac crest. Circumduction is identified by ipsilateral hip abduction and trunk and pelvic rotation during swing and vaulting is identified by contralateral ankle plantar-flexion during mid-stance (Winter, 1991). Locked-knee KAFO users adopt an individualized combination of these adaptations to compensate for reduced toe clearance during straight-leg gait

(Bowker et al., 1993; Irby, Bernhardt & Kaufman, 2007; McMillan, Kendrick, Michael, Aronson & Horton, 2004; Yakimovich, Kofman & Lemaire, 2005).

In a study by Zissimopoulos et al. (2007), healthy subjects walking with a locked KAFO showed patterns of pelvic obliquity, indicating hip hiking, and an increased thigh angle during swing phase, indicating circumduction. Several other sources have reported instances of irregular ankle and pelvic movements in conventional KAFO users (Herbert & Liggins, 2005; Irby et al., 2007; Irby et al., 2005). Abnormal movements associated with locked-knee KAFO gait contribute to the incidence of lower extremity joint and tissue dysfunction, which may be acquired by long-term users (McMillan et al., 2004).

2.3 Stance-Control Orthoses

Conventional KAFOs are effective in providing support to those with quadriceps weakness, but locked-knee gait may impede mobility when walking on inclined surfaces or stairs, manoeuvring around obstacles, or walking in conditions such as snow, gravel, or uneven terrain (Lemaire et al., 2009; McMillan et al., 2004). While some orthosis users may require the support of a locked KAFO, others may benefit from a device that offers support during weight bearing while allowing free knee motion for sufficient toe clearance during swing (Zissimopoulos et al., 2007).

Figure 4: Stance control orthosis (Horton's Stance Control, 2011).

Stance control orthoses (SCOs) have been designed to deliver these benefits. SCOs may be integrated into a KAFO (SCKAFO), shown in Figure 4.

One SCO design could be applied in a knee orthosis for those who do not require support at the ankle (Lemaire et al., 2009). SCKAFO manufacturers suggest a hip strength grade of 3 or greater to control leg position, ensuring that the knee is extended at terminal stance and to smoothly initiate swing phase (Bernhardt et al., 2006; Becker Orthopaedic, 2011). Research has shown that SCKAFO improve upon conventional locked knee KAFO by encouraging more normal gait patterns, improving mobility, reducing the energy cost of walking, and reducing compensatory strategies that may lead to chronic pain and loss of motion (Davis, Bach & Pereira, 2010; Herbert & Liggins, 2005; Irby et al., 2007; Irby et al, 2005; Irby et al. 1999; McMillan et al., 2004; Suga et al., 1998; Yakimoich et al., 2006; Yakimovich et al., 2005; Zissimopoulos et al., 2007).

2.3.1 Temporal spatial parameters

Throughout the literature there is a general agreement that, when compared to KAFO users, SCKAFO users exhibit temporal spatial parameters that more closely resemble the normal population (Davis et al., 2010; Irby et al., 2007; Irby et al, 2005; McMillan et al., 2004; Zissimopoulos et al., 2007). In 2007, Irby et al. (2007) found that KAFO users showed improvements in velocity, cadence, and stride length after six months of walking with a SCKAFO named the Dynamic Knee Brace System (Table 4). Other studies by Irby et al. (2005), Davis (2010), McMillan (2004), and Zissimopoulos (2007) agreed with the results in Table 4, reporting that velocity, cadence, stride length, and step length can increase when walking with a SCKAFO versus a locked-knee KAFO.

However, Herbert and Liggins (2005) revealed conflicting results in a single-subject study of a 61-year old man with post poliomyelitis. No significant differences in velocity, cadence, stride length or step length were found between stance-control KAFO and locked

Table 4: Temporal spatial parameters for 14 KAFO and SCKAFO users (Irby et al., 2007).

Temporal Spatial Parameter	KAFO	SCKAFO
Velocity (m/s)	0.72	0.82
Cadence (steps/min)	78	83
Stride Length (m)	1.07	1.17
Single limb support (percent per leg per cycle)	31	31

KAFO conditions. This discrepancy may be partially explained by the results from Irby et al. (2005) that compared walking patterns of a SCKAFO to a locked KAFO. Though novice KAFO users significantly increased velocity, cadence, and stride length when walking with a SCKAFO, the experienced users actually displayed slightly lower velocity and cadence. The authors suggested that experienced KAFO users may have ingrained gait strategies specific to ambulating with a KAFO that result in cautious gait strategies, including lowered velocity and stride length. Since the subject from the Herbert and Liggins (2005) study had 55 years of locked KAFO experience, he would most likely possess similar ingrained gait habits to those of the experienced users in the study by Irby et al. (2005).

Another factor that could explain the discrepancy between studies is the SCKAFO acclimation period. Irby et al. (2005) referred to only a brief orientation session with the SCKAFO and, though the subject in the Herbert and Liggins (2005) study received his stance-control KAFO six months prior to testing, no protocol was presented regarding the amount of time the participant spent practicing in stance control mode. The acclimation period has been shown to be an important factor in how effective SCOs are in improving gait patterns compared to locked-knee devices. Up to 6 months may be required for some users to make significant improvements in temporal spatial parameters (Irby et al. 2007).

The subject's hip flexor strength may also be partially responsible for the inconsistency presented in the literature. In the Irby et al. (2005) study, experienced users were shown to decrease velocity and cadence but were also shown to have lower hip flexor strength and significantly lower hip and knee extensor strength than their novice counterparts. Participants in the Herbert and Liggins (2005) study also had lower hip strength grades in terms of flexion and adduction than typically recommended for SCKAFO use. Lower hip muscle strength limits the user's ability to control the orthosis and may result in a protective gait strategy (Bernhardt et al., 2006; Becker Orthopaedic, 2011).

2.3.2 Kinematics

Arguably, the most important gait improvement when walking with a SCKAFO versus a locked-knee KAFO is the decrease in compensatory movements related to straight-knee gait. Studies have reported that SCKAFOs have marginal effects on normal gait patterns of able-bodied subjects and improved gait kinematics for KAFO users (Herbert & Liggins, 2005; Yakimovich et al., 2005; Yakimoich et al., 2006; Zissimopoulos et al., 2007). Though these improvements may take time to develop, the decreased instance of compensatory strategies may help users maintain better mobility and range of motion after prolonged SCKAFO use than would be possible with a conventional KAFO (Irby et al. 2007; McMillan et al., 2004).

While a lack of knee flexion during gait is an obvious result of walking with a locked-knee KAFO, straight legged gait during stance is also typical for people with quadriceps weakness who don't use an orthosis (Lohmann Siegel, Stanhope & Caldwell, 1993). SCKAFO use can significantly increase peak knee flexion over time to produce a near normal knee flexion curve during swing phase (Herbert & Liggins, 2005; Irby et al. 1999; Irby et al. 2007).

As previously stated, one of the most common gait adaptations related to KAFO use is hip hiking. Zissimpoloulous et al., 2007 (Figure 5) showed that walking in unlocked and auto SCKAFO modes resulted in significantly less pelvic obliquity, compared with locked-knee mode. Other research agreeed with these findings, since SCKAFO interventions reduced extraneous pelvic movements, decreased vertical excursion at the pelvis, reduced pelvic obliquity, and generated smoother, more symmetrical pelvic movements (Herbert & Liggins, 2005; Irby et al. 2007; McMillan et al., 2004; Yakimoich et al., 2006).

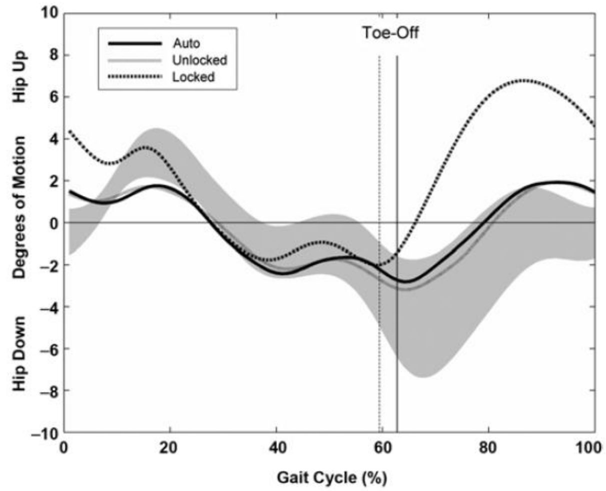


Figure 5: Pelvic obliquity when walking in auto, unlocked, and locked SCKAFO modes (shaded region is SD for able-bodied people) (Zissimopoulos et al., 2007).

Circumduction is another adaptation made at the hip, which is identified by ipsilateral hip abduction and trunk and pelvic rotation. Research by Yakimovich et al. (Yakimovich et al., 2005) reported that, compared to a conventional KAFO, walking with a SCKAFO resulted in a decrease in hip abduction angle abnormalities. Similar findings have reported that SCKAFO use decreased extraneous hip and trunk movements and had a noticeable effect on pelvic rotation, suggesting a more efficient gait pattern (McMillan et al., 2004; Herbert & Liggins, 2005).

Vaulting is the third of the three common gait strategies associated with conventional KAFO use. Research by Irby et al. (2005) and Herbert and Liggins (2005) found that walking with a SCKAFO reduced the instance of vaulting in KAFO users.

Interestingly, the 2007 study by Irby et al. reported no improvements in compensatory strategies between KAFO and SCO conditions. The authors suggested that, since KAFO and SCKAFO users adopted an individualized combination of compensatory strategies specific to their weaknesses, participants may make significant improvements to their individual gait patterns even though significant results are not apparent in grouped data. This explanation may also apply to other studies involving multiple-subjects.

2.4 Design Overview for Stance Control Orthoses

2.4.1 Design considerations

In recent years, improvements to lower limb orthoses have resulted in lighter, stronger, more durable, and better fitting devices (Esquenazi & Talaty, 2000). However, design issues still exist that may frustrate users to the point that they abandon their assistive device (Kaufman et al, 2001; Peethambaran, 2000; Phillips & Zhao, 1993; Yakimovich et al., 2009). Ideally, a SCKAFO design will be highly functional, reliable, comfortable, lightweight, inexpensive, and easy to use while maintaining its aesthetic appeal and creating minimal noise or interference during activities of daily living (Bernhardt et al., 2006; Herbert, 2006; Yakimovich et al., 2005; Meyer, 1974).

In addition to these factors, Yakimovich et al. (Bernhardt et al., 2006) outlined a list of additional features that are advantageous in a SCKAFO design. The list included locking of the knee or resisting knee flexion at any knee or ankle angle (not only when knee is fully extended); unlocking the knee at any knee or ankle angle when the limb is unloaded; controlled knee flexion during stance; smooth transition between stance and swing modes; assisting knee extension during stance; and switching between stance and swing modes without requiring a knee extension moment to unload the joint. The joint must also have a quick reaction time and function should not be sacrificed when presented with a fast cadence. These additional features would be advantageous during tasks such as sitting and stair climbing and may be beneficial for users when regaining balance after a stumble. They would also promote a smooth gait progression and a close to normal gait pattern. Essentially, an ideal orthosis would provide the user with complete support during the stance phase of gait, without restricting joint movements during swing (Kaufman et al, 2001). Engineers are presented with a challenge when attempting to incorporate all of these features within a single SCKAFO design.

2.4.2 Design categories

Many stance control approaches exist and each presents certain advantages and disadvantages that affect performance and user satisfaction. An article by Lemaire et al. (2009) outlined four main categories of SCKAFO design, two of which use mechanical stance control methods. The first mechanical design relies on limb position to initiate device engagement. At a certain point in the gait cycle, such as foot strike or toe off, a spring loaded or weighted prawl is activated to lock the knee in extension. While these orthoses tend to be the most cosmetically appealing, some users may feel they are not well supported by the small, lightweight frame (Yakimovich et al., 2009). The major issue associated with these SCKAFOs is that they only lock when the leg is in full extension and therefore will not provide support in situations when the knee is already flexed, such as stair climbing or when recovering from a stumble. The other type of mechanically controlled SCKAFO addresses this issue. A mechanism at the footplate detects weight-bearing to initiate stance control mode. This design allows knee-flexion resistance at any knee angle; however, the use of mechanisms external to the knee joint can result in a bulky, heavy orthosis that some users may not tolerate (Lemaire et al., 2009). The electronic stance control method is the third design. This approach also permits knee flexion resistance at any knee angle and uses pressure sensors to recognize the weight-bearing stage and switch between stance-swing modes. Electronic control can improve joint function and reliability, but batteries add additional bulk, weight, and maintenance (Irby et al, 2005; Yakimovich et al., 2009; Irby et al. 1999). The last stance control approach outlined in the article uses varied knee flexion resistance states to achieve stance control based on pre-set knee angles and predetermined relationships. This approach may not be a reliable source of support in uncommon situations, such as a stumble (Lemaire et al., 2009).

2.4.3 Angular velocity control approach

As discussed in the previous section, stance-control orthoses are typically designed to respond to changes in knee angle, ankle angles, or ground reaction force. Engineers and researchers at The Ottawa Hospital Rehabilitation Centre (TOHRC) have developed a new angular velocity-based control method, named the Ottawalk-Speed (OWS). This approach does not require the use of sensors or actuators external to the knee joint, and therefore has the capacity to function within a compact knee unit (Lemaire et al., 2009). A compact unit may be especially beneficial for people with limited or temporary knee extensor weakness who do not require a full SCKAFO.

This novel approach is based on the idea that, during a knee collapse event such as a stumble or fall, knee-flexion angular velocity is greater than that during walking. The design criteria as stated by Lemaire et al. (2009) are as follows:

- Resist knee flexion when engaged but allow knee extension
- Permit free knee motion when not engaged
- Function without failure at high loading conditions (i.e. fast cadence, ascending/descending stairs, falling) with capability of resisting a 154 Nm knee-flexion moment (estimated for 90 kg user in stair ascent)
- Allow easy adjustment and integration with standard orthotic components
- Minimal frontal-plane dimensions to maintain cosmesis and limit occurrence of bilateral joints touching each other on medial side
- Maintain low noise level
- Control system internal to knee joint

This approach uses hydraulic technology to detect knee-flexion angular velocity. Flexion control engages when knee flexion angular velocity surpasses a set threshold. As the knee flexes, fluid within the joint's flexion chamber is pushed through a valve into the extension chamber. A spring force acts upon the valve to create the threshold velocity, which can be adjusted to suit individual users. As the knee continues to flex, the moving fluid creates a drag force on the valve to oppose the preset spring force. As long as the drag force does not meet or exceed the spring force, the valve will remain open. Therefore, as long as knee flexion velocity does not exceed the preset angular velocity threshold, the valve stays open. Once knee flexion velocity reaches or exceeds the threshold, the valve will close. This blocks fluid flow through the internal joint chambers and subsequently causes the joint to resist knee flexion. This device has shown promise in initial testing and offers the security of knee-flexion resistance when the knee is flexed while maintaining a lightweight, aesthetic, and compact SCO design (Figure 6). However, the angular velocity control design must first be tested on a target population to assess the functional and biomechanical performance, safety, reliability and user satisfaction before it can be used by consumers.

During mechanical testing by Lemaire et al. (2013), the joint engaged within 2 deg knee flexion and resisted moment loads of over 150 Nm at the lowest angular velocity threshold, and completed 400,000 loading cycles without mechanical failure or wear that would affect function. Functional testing also revealed that able-bodied walking patterns with the OWS were similar to normal when walking below the threshold velocity and, if the joint engaged when purposely walking at speeds above the threshold velocity, the OWS did not cause unsafe gait patterns. Knee collapse tests revealed that the device successfully resisted and stopped knee flexion with peak knee moments of up to 235.6 N.

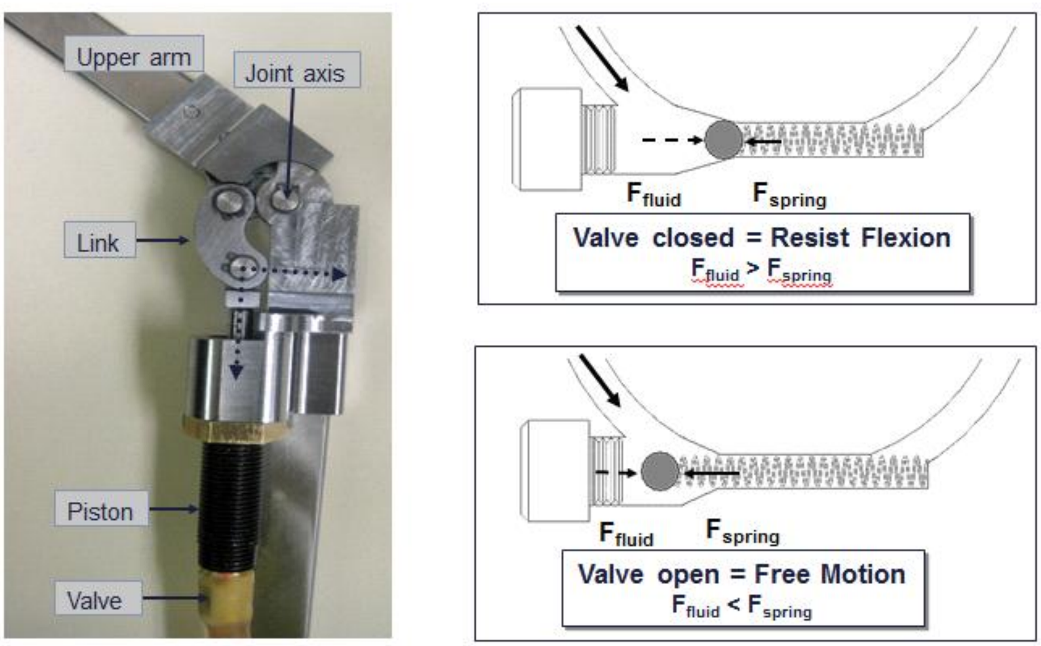


Figure 6: Ottawalk Speed (OWS).

2.5 Inducing a Stumble

Falling is an everyday risk that can be detrimental to well-being. Various populations at increased risk of falling, including older adults and people with quadriceps weakness, may develop debilitating anxiety, loss of mobility and independence, and could incur serious injury or die from fall incidents (Lord, Menz & Tiedemann, 2003; Perell et al., 2001). Induced trip and slip methods have been previously developed and are important for fall prevention programs, and for researchers studying the underlying mechanisms related to fall risk (Pavol, Runtz, Edwards & Pai, 2002; Cordero et al., 2003; Pai & Bhatt, 2007). For our purposes, an induced stumble method was used to evaluate OWS stance control orthosis.

Induced stumble methods should be safe and repeatable in their application, and should mimic a real-world stumble situation. Outcomes from these methods are impacted by factors including gait speed, perturbation type, and time of the perturbation during the gait cycle. (Cordero et al., 2003; Smeesters et al., 2001). Previously established perturbation methods include:

- **Gait obstruction**: A rope or cable is attached to the leg, restricting forward motion at different points of the gait phase (Forner Cordero, Koopman & van der Helm, 2003; Delmarcelle et al., 1997; Smeesters, Hayes & McMahon 2001). Alternatively, hidden obstacles can be extruded from the floor, or obstacles dropped into the path, to obstruct toe clearance during swing (Bieryla, Madigan & Nussbaum, 2007; Eng, Winter & Patla, 1994; Grabiner, Koh, Lundin & Jahnigen, 1993; Schillings, Van Wezel & Duysens; 1996).

- **Translating surfaces**: Rapid surface translation in stance via sliding devices embedded in the ground (Pavol et al., 2002; Pai & Bhatt, 2007; Parijat & Lockhart, 2011; Tang, Woollacott & Chong, 1998).

- **Treadmill acceleration/decelerations**: Rapid treadmill acceleration, or a combination of both acceleration and deceleration, during standing or gait (Dietz, Quintern & Sillem, 1987; Owings, Pavol & Grabiner, 2001; Vilensky, Cook & Cooper, 1999). Overhead bodyweight support harnesses ensure safety.

While these methods have been shown to induce stumbles in repeated trials, some methods that use mechanical objects to induce trips and slips utilize minimal safety structures, such as safety mats or a spotter. These safety measures may not be as effective as a safety harness for preventing injury in the event of a fall. Additionally, some trip or slip perturbations may be anticipated after repeated trials, and attempts to prevent anticipation by obstructing vision may not mimic real-life stumble situations. On the other hand, treadmill perturbations that induce a stumble can be easily randomized, may be less likely to be anticipated by participants during trials, do not require other mechanical objects to apply the perturbation, and can use overhead safety harnesses to ensure safety in the event of a fall.

From the literature, a few reports are of particular interest in the development of a stumble method that will induce knee flexion upon landing. One stumble method, reported by Vilensky et al. (1999), used a treadmill reversal perturbation to cause a stumble in participants. During gait, the treadmill belts were rapidly decelerated and reversed, followed by a rapid acceleration of the belts back to the starting speed. The perturbation was applied at foot strike, and lasted between 230-350ms. The resulting recovery strategy included rapid flexion of the knee on the stance limb.

Another study by Tang et al. (1998) applied a slip perturbation to young adult participants via a translating surface, and found that the key to regaining stability within one gait cycle was the coordination between the two legs, as well as early, prolonged activation of bilateral anterior leg muscles and anterior and posterior thigh muscles. Since anterior thigh muscles responsible for knee extension are important for regaining stability after a perturbation, people with knee extensor weakness may not recover as easily as able-bodied persons since the ability to extend the knee is so important to preventing knee collapse (Lohmann Siegel, Kepple & Stanhope, 2005) For these reasons, this perturbation strategy should be useful for testing the OWS's ability to prevent orthosis users from falling in stumble situations.

The CAREN Extended virtual reality system can be used in rehabilitation and research related to posture, balance and post-traumatic gait training, and offers a safe and repeatable means of applying perturbations that cannot be achieved in a real world setting. Since CAREN has the capacity to perturb the virtual environment without the use of mechanical methods, it may mimic a stumble situation more realistically than typical laboratory settings, making it an ideal environment for stumble development and application of a stumble method.

2.6 Summary

SCKAFOs improve upon conventional locked knee KAFO by reducing compensatory strategies that may lead to chronic pain and loss of motion. The novel OWS SCKAFO has the potential to improve upon bulky and heavy SCKAFOs currently on the market with its lightweight design and ability to engage and prevent knee flexion in unpredictable knee collapse situations. However, the angular velocity control design must be evaluated on the target population of orthosis users before release to the consumer market. Evaluation of normal walking, different surfaces (treadmill level, incline, decline, rolling hills, uneven), and every day stumble situations would improve our understanding of angular velocity control as an orthosis control mechanism. The CAREN-Extended system is an ideal environment for OWS SCKAFO testing. If a repeatable and controlled perturbation method can be created, this tool can simulate stumble situations and different real world walking surfaces within a safe setting.

Chapter 3. Methods

3.1 Instrumentation

3.1.1 CAREN-Extended

CAREN-Extended is a virtual reality environment that aims to improve movement research and rehabilitation practices, especially in the areas of posture and balance (Fung, Malouin, McFadyen, Comeau, Lamontangne, Chapdelaine, Beaudoin, Laurendau, Hughey & Richards, 2004; Kruger, Bell & Schnall, 2009; Lees, Vanrenterghem, Barton & Lake, 2007; Oates, Perez & Fung, 2008). This technology allows operators to manipulate platform and treadmill movements to perturb and test users in a safe and repeatable manner (McAndrew, Dingwell & Wilken, 2010). Figure 7 displays the CAREN-Extended setup.

The CAREN-Extended system contains a 180°, concave virtual projection screen, supplied by four overhead projectors. The five-meter screen allows the displayed image to extend beyond the subject's periphery for nearly complete visual immersion. A circular hydraulic motion platform (3m diameter) capable of movement in six degrees of freedom (DOF) is situated within a 5m cylindrical pit. Barton (Barton, Vanrenterghem, Lees & Lake, 2005) described platform movements as three translational displacements (sway, surge, and heave) and three rotational displacements (pitch, roll, rotation). A dual-track treadmill is embedded in the motion platform, and each belt is instrumented with Bertec (Bertec, Ohio, U.S.A.) force plates. The dual-track treadmill belts can operate with linked or independent speeds, and can function in either constant velocity or self-paced modes. A 12 camera Vicon motion capture system (Vicon, Oxford, England) borders the platform, along with a five speaker integrated surround sound system that synchonizes with visual applications to enhance user perception of the virtual environment.

Safety features include a retractable transfer bridge and an overhead safety structure that supports a fall-arrest harness. Mandatory safety procedures demand that users are secured to the harness while immersed in the environment. ‘Emergency Stop’ and ‘Suspend’ buttons can be accessed by operators at any time. Activating the emergency stop causes the treadmill and platform to abruptly stop in the current position. Activating suspend causes the treadmill to gradually stop and the platform to return to its home position. A safety light beam is located at the back of the treadmill. If the user moves too far to the back of the treadmill, the light beam is broken and the system immediately suspends.

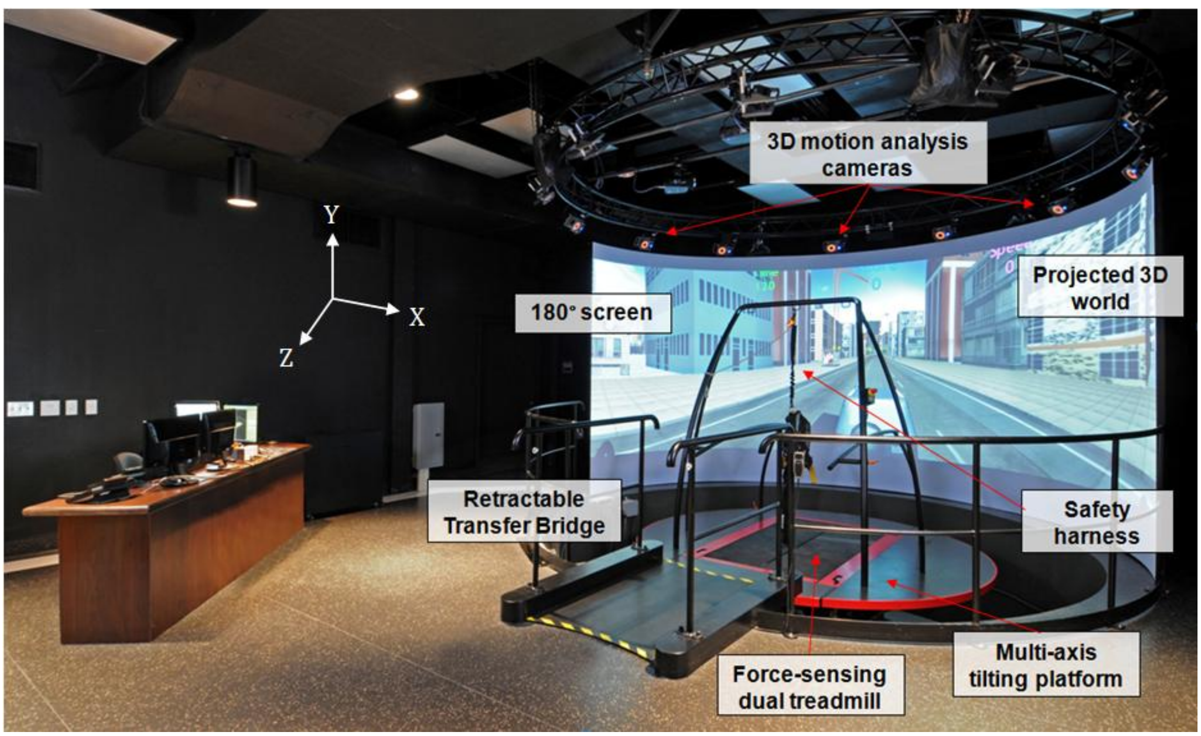


Figure 7: CAREN-Extended at TOHRC.

The CAREN system uses D-Flow control software to synchronize hardware components. Programmers can create and adapt applications to simulate case-specific virtual scenarios (Makssoud, Richard & Comeau, 2009). Applications settings for treadmill and platform motion may run as a preset program (useful for research practices that must be precise and repeatable) or may be manually manipulated during trials by the operator (useful in clinical settings for individualized rehabilitation programs). D-Flow can use reflective marker data from Vicon to manipulate the environment in real time (i.e. self paced mode).

The self-paced treadmill mode allows users to move at their self-selected speed within the virtual environment, with speed changes automatically accommodated by the VR system. D-Flow uses real time feedback from two or more Vicon markers to calculate the user's position in the treadmill longitudinal plane (Z-plane), and treadmill speed is controlled based on marker movement from the starting position. The treadmill accelerates as the person moves forward on the treadmill and decelerates as the person moves backward, keeping the person near the middle of the treadmill.

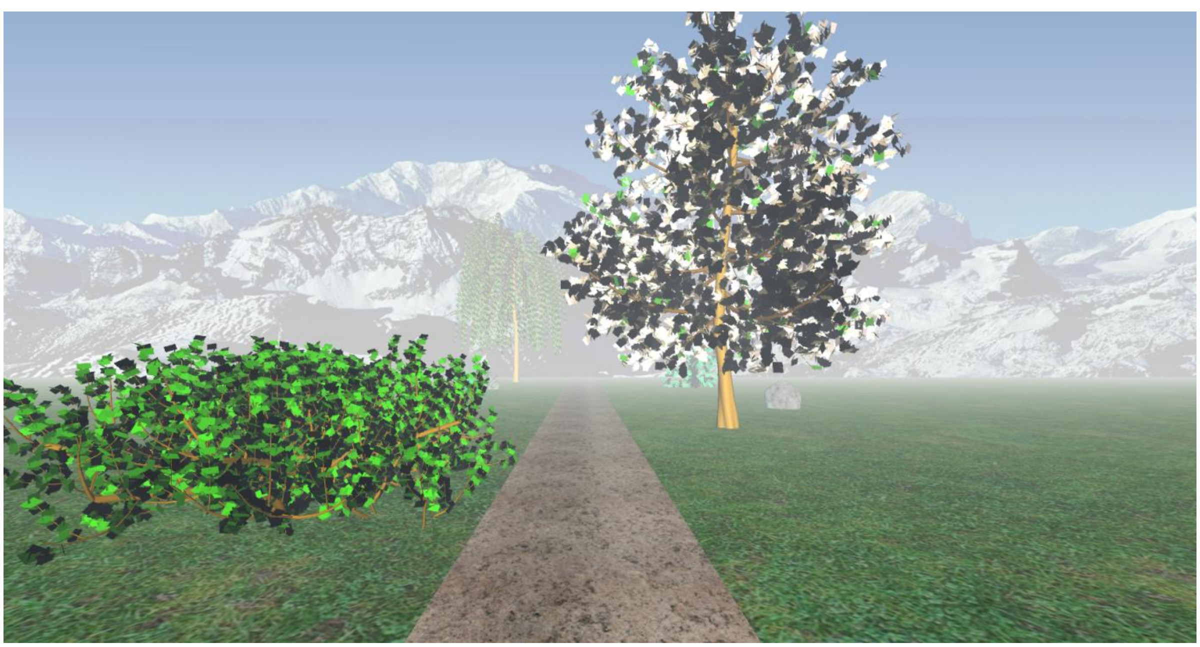


Figure 8: Screenshot of "The Park" virtual scenario for CAREN-Extended.

“The Park” is a custom designed virtual application for the CAREN-Extended system that provides a path through a grassy park setting, containing numerous trees, rocks, and a mountainous backdrop. Each scene consists of twenty-metre sections that provide treadmill level, uphill and downhill slopes, rolling hills, and uneven conditions. These conditions can be altered in real time to change incline and intensity, to accommodate the user’s functional level. A series of sections can also be assembled as a preset scene.

Uphill and downhill slope conditions were set to 5° (1:11.43 slope/8.75% grade) for OWS testing trials, to correspond to typical 1:12 slope access ramps (Lemaire, O’Neill, Desrosiers & Robertson, 2010).

3.1.2 Vicon motion capture

A 12-camera Vicon Motion Capture system (Vicon, Oxford, England) is synchronized with CAREN for kinematic data collection (Figure 9). This system uses infrared cameras to track passive reflective markers in 3D space using an intricate mapping system (Vicon, 2011). By placing reflective markers at anatomical locations, limb segment movements are tracked and trials can be processed to measure segment and joint kinematics. Vicon data can also be used to calculate temporal-spatial parameters.

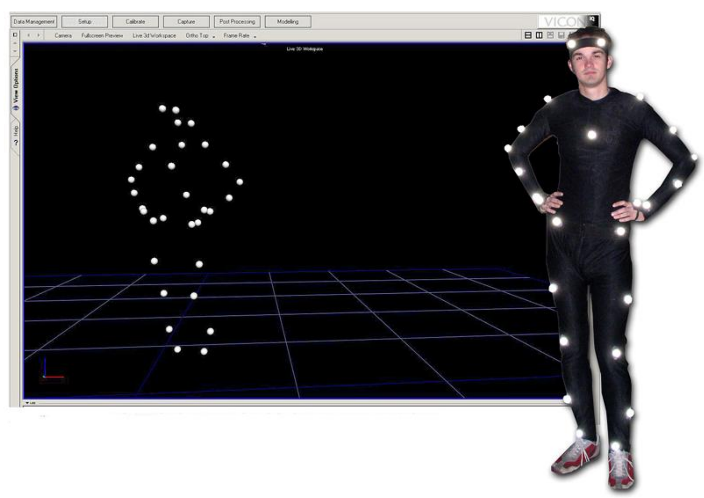


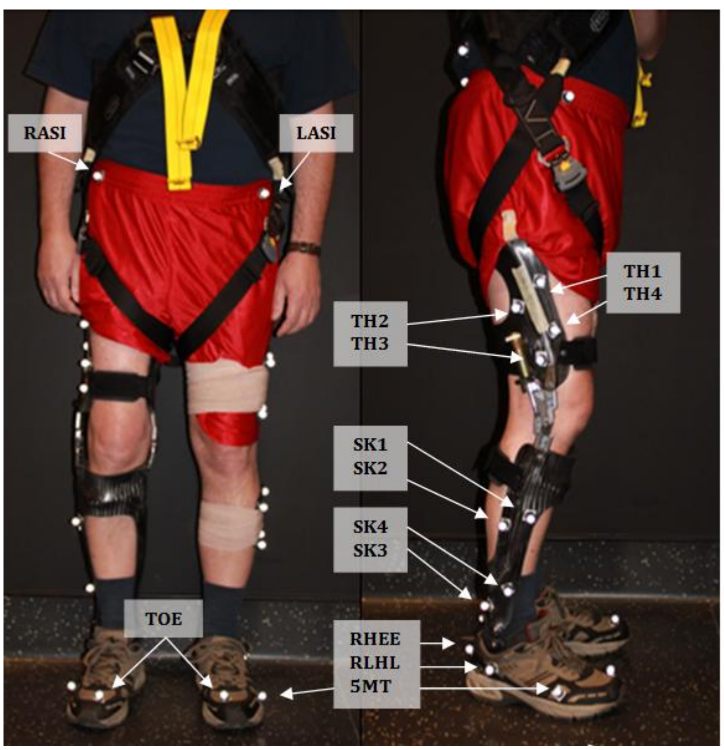
Figure 9: Vicon subject and software (Vicon, 2011).

3.2 Questionnaire

A questionnaire was administered to OWS participants following testing (Appendix F). This questionnaire was based on a previously used questionnaire for KAFO assessment (Bernhardt et al., Peethambaran, 2000). Participants assessed the OWS SCKAFO on factors shown to influence user satisfaction towards orthoses; including, comfort, cosmesis, weight, noise during movement, ease of use and application, and feeling of stability.

3.3 Marker set

A lower body 6 DOF marker set was used for all kinematic data collection (i.e., stumble method and OWS evaluation) (Figure 10). Thirty reflective markers were placed bilaterally at the anterior iliac spine, posterior iliac spine, lateral pelvis, thigh, shank, toe, 5th metatarsal, heel and lateral heel. For able-bodied participants, thigh and shank plates, each



Anterior superior iliac spine:	LASI, RASI
Posterior superior iliac spine:	LPSI, RPSI
Hip reference:	LASI_2, RASI_2
Thigh plate:	LTH1, LTH2,LTH3, LTH4, RTH1,RTH2, RTH3, RTH4
Shank plate:	LSK1, LSK2, LSK3, LSK4, RSK1, RSK2, RSK3, RSK4
Toe:	LTOE, RTOE
5th metatarsal:	L5MT, R5MT
Lateral heel:	LLHL, RLHL
Heel:	LHEE, RHEE

Figure 10: Lower body 6 DOF marker set for OWS kinematic data collection (OWS user).

with four markers, were attached to each leg. For OWS testing, thigh and shank plates were attached to the non-braced leg and four thigh and four shank markers were manually placed on the orthosis of the braced leg.

In addition to the reflective markers, eight anatomical landmarks were defined bilaterally at the medial and lateral malleoli and medial and lateral femoral epicondyles using a digitizing wand. This practice is useful in reducing the instances of lost markers. These anatomical markers were defined during a short motion trial, called a digitizing trial, by touching the digitizing wand to the respective anatomical landmark location. In post processing, Visual 3D recognizes the anatomical marker when the wand tip is depressed. Figure 11 illustrates clips of a digitizing trial in Visual 3D.

For stumble development and OWS testing, four pelvis markers were identified in D-Flow as reference markers for self-paced mode (left and right anterior and posterior superior iliac spine). All participants were instructed to wear snug-fitting, non-reflective clothing and running shoes for their RVR Lab session.

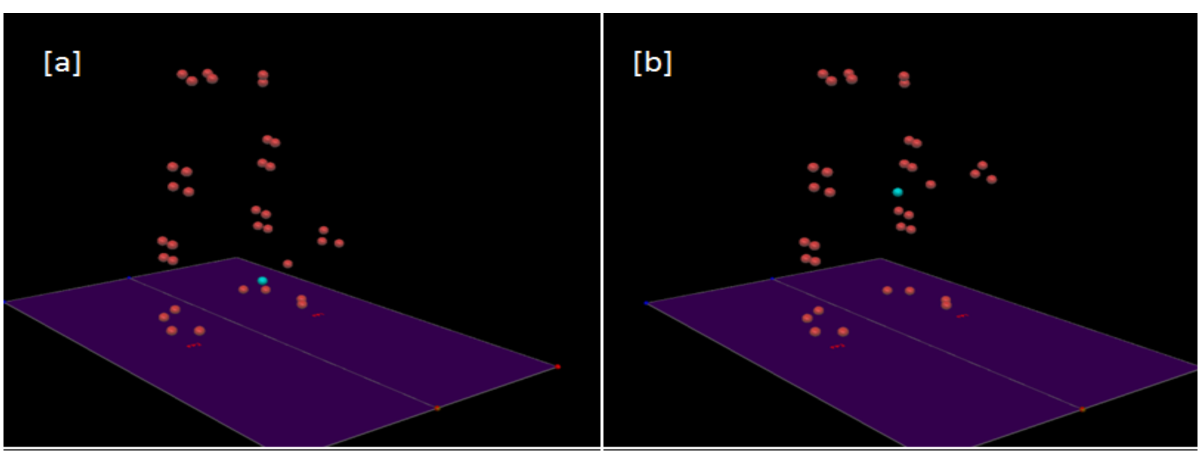


Figure 11: Lower body 6 DOF marker set and digitizing wand during digitize trial. The wand defines anatomical landmarks on the left side of the body by touching: [a] medial malleolus, [b] medial femoral epicondyle.

3.4 Manual muscle testing

Manual muscle testing was performed on all OWS evaluation participants. Table 5 outlines the Manual Muscle Testing (MMT) grading scale, as adapted from Kendall, McCready & Provance (1993). This standardized test is a common and reliable method of determining muscle strength (Dufree & Iazzo, 2006). MMT tests were performed on all potential candidates recruited for OWS testing. The same orthotist conducted all MMT tests to ensure consistency of strength grades between subjects.

Table 5: Manual Muscle Testing Criteria (Kendall, McCready & Provance, 1993).

MMT Isometric Grading		
5	Normal	Holds test position against maximal resistance
4+	Good +	Holds test position against moderate to strong pressure
4	Good	Holds test position against moderate resistance
4-	Good -	Holds test position against slight to moderate pressure
3+	Fair +	Holds test position against slight resistance
3	Fair	Holds test position against gravity
3-	Fair -	Gradual release from test position
2+	Poor +	Moves through partial ROM against gravity or moves through complete ROM gravity eliminated and holds against pressure
2	Poor	Able to move through full ROM gravity eliminated
2-	Poor -	Moves through partial ROM gravity eliminated
1	Trace	No visible movement: Palpable or observable tendon prominence/flicker contraction
0	0	No palpable or observable muscle contraction

3.5 Stumble Method

3.5.1 Preliminary testing

A series of stumble generation methods were qualitatively pilot tested in the RVR lab. These methods were applied at different times during gait, at a constant self-selected treadmill speed:

- **Rapid translations**: A rapid forward or downward platform translation at foot strike (FS) produced a mild stumble that was easily recovered in a few shortened, widened steps. A large platform displacement was required (> 35cm).
- **Platform pitch:** Rapid platform incline/decline after FS was somewhat effective for stumbling onto a bent knee. However, a platform pitching method may not be appropriate for all walking conditions, such as uphill, downhill or rolling hills.
- **Treadmill acceleration/deceleration**: Quickly decelerating linked treadmill belts during swing, followed by quick acceleration to the original speed after FS, caused a repeatable stumble with knee flexion landing.
- **Split-speed**: Using self-paced mode in the level condition, one treadmill belt sped up while the other slowed down at mid swing. This method worked for any platform angle, accommodated self-paced mode, isolated the stumble to a predetermined leg, was repeated in several trials, and caused bent-knee landing.

3.5.2 Perturbation Description

The split-speed treadmill perturbation consisted of one treadmill belt speeding up while the other slowed down. The perturbation was applied at mid-swing so that the ipsilateral treadmill belt could accelerate to reach the higher speed at FS of the perturbed step. System settings allow either a ‘left trip’ or ‘right trip’ event to be applied (i.e., landing on the left or right leg). If a left trip was applied, the perturbation was applied to the right foot; therefore, the perturbation was triggered when the right leg was in mid-swing and caused the right treadmill belt to speed up. To ensure correct timing and consistency, mid-swing was determined from toe marker positions. Two virtual objects created in the D-Flow software represented left and right toe markers. The markers were hidden so that they were not visible in the virtual scene. When an “Enable Trip” event was applied, the system recognized a collision event when the two toe objects passed each other in mid-swing. The virtual collision triggered the treadmill perturbation. The perturbed leg must be moving forward for a collision to occur, to ensure that the appropriate leg is perturbed.

During the perturbation, belt speeds were split based on a percentage of the current treadmill speed. On a scale of 0-1, a trip speed of zero represented the greatest difference in belt speeds (i.e., one belt accelerates to twice the current speed and the other belt is stopped). The treadmill belt perturbation was programmed as a square wave of period 0.5s. The actual speed that the treadmill attained was influenced by maximum treadmill acceleration ($1.5 m/s^2$) and speed (2.5 m/s) limits.

3.5.3 Participants

A convenience sample of five healthy, able-bodied volunteers participated in this study. Participants were emailed a recruitment letter via TOHRC internal email, and provided informed consent (Appendix A). This study was approved by The Ottawa Hospital and University of Ottawa Research Ethics Boards.

3.5.4 Experimental procedures

After reviewing the protocol and “Patient CAREN Orientation and Procedure Form” (Appendix B), and participants provided written consent as per the Pilot Information and Consent Form (Appendix C). A research assistant fit a safety harness to the participant and attached the lower body 6 DOF marker set to the skin, clothing and safety harness for kinematic data collection. Participants were oriented to the CAREN-Extended system and all safety procedures were reviewed before being escorted to the motion platform by the research assistant and secured to the overhead fall-arrest structure with a safety tether. A dynamic subject calibration was completed prior to testing to ensure good marker tracking by D-Flow throughout trials, which was necessary due to the use of self paced mode. Participants then completed two 160m long, treadmill level accommodation trials in “The Park” application, one at constant treadmill speed and one with self-paced mode. One static and two digitizing trials were collected before stumble trials commenced for use in post-processing. In the event that a marker dropped during the session, an additional static trial was collected.

For each stumble trial, participants walked along a 100m, level path in “The Park”, using self-paced treadmill mode. Participants were informed that a stumble-inducing perturbation would be randomly applied during each trial. Participants were able to rest between trials as needed. Trials were repeated until ten stumbles were collected or until session time reached one hour. Participants were given breaks as needed. Due to time restrictions related to accommodation time for self-paced mode, only three stumbles were completed for participant five. Participant five agreed to return for a second session where all ten stumbles were completed. Only seven stumbles were completed for participant one due to time restrictions.

Kinematic data were collected by 12 Vicon motion capture cameras, sampled at 120Hz. For each trial, D-Flow recorded the time elapsed, platform position, treadmill speed, and distance and split speed perturbation events. Manual records were kept for each trial relating to which leg was perturbed, the tile when the perturbation was applied, and notes relating to the participant's recovery response to the perturbation. Digital video was recorded for all stumble trials during the session.

3.5.5 Data analysis

Marker data were cropped using Vicon Nexus to include three gait strides preceding the perturbation and three strides following the perturbation. After exporting to Visual 3D (C-Motion Inc., Germantown, MD) marker data were filtered with a dual pass 10Hz Butterworth filter and gait events were identified. TL gait values for each trial were derived from the three strides before the perturbation. Foot strike (FS) and foot off (FO) events were identified for TL strides. Events representing the stance phases following the perturbation were created at FS and FO of the perturbation step (PERTFS, PERTFO), recovery step 1 (REC1FS, REC1FO), and recovery step 2 (REC2FS, REC2FO). For example, a 'left trip' included a perturbation step on right leg, recovery step 1 on the left leg, and recovery step 2 on the right leg. For a 'right trip', these events occurred on the opposite leg.

Ankle, knee, hip, and pelvis angles and angular velocities were extracted for TL, perturbation, and recovery events. The independent variable was the gait condition (TL or perturbed). Dependent variables included joint angles (ankle, knee, hip, pelvis), joint angular velocities (ankle, knee, hip), and temporal spatial parameters (step length, step width, double support time (DST), stance time, swing time, stride time). Maximum and minimum values in stance and swing were extracted for TL, perturbation, and recovery strides. The perturbation stride was from PERTFS to REC2FS and recovery 1 stride was from pre-stumble FO to REC1FO. Differences in temporal spatial, angle, and angular velocity values for perturbed strides were compared with TL strides using descriptive statistics. TL values for left and right legs were averaged. Manual data and digital videos were reviewed and treadmill speed from D-Flow was examined to determine walking speed changes during perturbation recovery.

3.6 OWS Evaluation

3.6.1 Participants

Potential participants were community ambulators, identified by orthotists within their circle of care at TOHRC. Recruitment letters were sent by mail by the research assistant (Appendix D). Inclusion criteria required that eligible candidates had impaired knee extensor strength with hip flexor strength of at least 3 (Becker Orthopaedic, 2011), as determined by the orthotist using the MMT Scale; had the capacity to extend their limb forward during the swing phase without aid; were able to walk continuously for 100m; were able to walk on in incline and decline conditions; and did not possess significant pain, loss of motion, joint dysfunction, or any other medically relevant condition that would affect the ability to safely complete testing. Those who met the inclusion criteria would be considered at the higher end of the KAFO/SCKAFO user population in terms of muscle strength and ambulatory ability, but still required an orthosis for unpredictable situations, such as a stumble.

3.6.2 Experimental procedures

Three sessions were completed on three separate days: casting, fitting and training, and OWS testing. Casting and fitting and training sessions took place at the Prosthetics and Orthotics Service (P&O) while OWS testing was completed in the RVR Lab at TOHRC.

3.6.2.1 Casting session

Potential participants met with the research assistant at P&O to review the study protocol and provided written consent as per the OWS testing information and consent form

(Appendix E). Information including sex, age and weight, along with a brief history of orthosis, use was obtained and recorded on the Participant Information Sheet (Appendix F). To ensure that recruited persons met study specifications, the orthotist performed Manual Muscle Testing on hip flexors and extensors, knee flexors and extensors, and ankle dorsiflexors plantar flexors. MMT grades were recorded by the research assistant.

An orthotist took a negative cast of the braced (affected) leg of each potential participant. The cast was used to fabricate custom orthoses for each of the three eligible participants. A more detailed description of the casting and fabrication process is described in Appendix G. Casting sessions were approximately one hour in length.

3.6.2.2 Fitting and training session

Once OWS SCKAFO fabrication was complete, participants returned to P&O, and met with the research assistant for a brief review the study protocol and to verbally confirm consent. The orthotist assisted the participant with donning the OWS SCKAFO, and made necessary adjustments to ensure proper fit. Participants were then escorted to parallel bars to practice walking with the orthosis At this time, the OWS threshold was adjusted by tightening or loosening spring resistance on the control valve to set the velocity angular velocity threshold slightly above the maximum angular velocity during terminal stance, so that the joint would not engage during walking. Once participants were comfortable walking, they practiced several knee collapse events within parallel bars by transferring all their weight to the braced leg, allowing the knee to flex freely and quickly without resistance, and allowing the orthosis to engage to stop them from falling. This was similar to training performed in a previous study, with able-bodied participants wearing OWS SCKAFOs (Lemaire, Samadi, Goudreau & Kofman, 2012). Knee collapse events were repeated until participants were able to actively engage the OWS, and felt comfortable in the joint's capacity to prevent them from falling. Training continued until both the participant and orthotist were confident in the participant's ability to manoeuvre the OWS. Each fitting and training session was approximately one hour in length. Participants were given a copy of the Patient CAREN Orientation and Procedure Form to review at home before returning to TOHRC for OWS testing in the RVR Lab.

3.6.2.3 OWS evaluation

Participants met with the research assistant for a brief review the study protocol and to verbally confirm consent before preparing for testing in the RVR Lab. The research assistant aided participants with donning the OWS SCKAFO, if needed. After the OWS engagement threshold was adjusted, the participant let their braced limb collapse to ensure that the device would engage appropriately, similar to the training session. The participant was fit to an overhead harness and the lower body 6 DOF marker set was attached to the skin, clothing, harness, and OWS orthosis for kinematic data collection. Participants were oriented to the CAREN-Extended system, and all safety procedures were reviewed before being escorted to the motion platform by the research assistant, and secured to the overhead fall-arrest structure with a safety tether.

A dynamic subject calibration was completed prior to testing to ensure good marker tracking. Participants then completed several accommodation trials of "The Park" application to adjust to the virtual environment. Accommodation trials did not exceed 80m. First was a TL trial at a constant treadmill speed adjusted by the operator to match what the participant felt was their typical walking speed, followed by a TL self paced trial, where the participant learned to control the treadmill speed via his position on the platform.

Finally, participants accommodated to uphill, downhill, rolling hill and uneven ground conditions (separated by TL conditions) in self paced mode. One static and two digitizing trials were collected before stumble trials commenced. An additional static trial was collected in the event that a marker dropped.

OWS testing consisted of five 100m walking trials in “The Park” in self paced mode. All trials began and ended with a TL condition and contained three of uphill, downhill, rolling hills or uneven simulated surface conditions. Participants were informed that stumble-inducing perturbations would be applied during trials, but did not know when the perturbations would be applied. Since all participants were wore their orthosis on the right leg, ‘right trip’ perturbations were applied. The stumble perturbation described in 3.2.2 was randomly applied at a point along the path during TL walking. Participants were able to rest between trials as needed. Trials were repeated until five trials were collected or until session time reached three hours. The operator adjusted the treadmill perturbation length (s) as needed to accommodate individual walking speeds. One participant repeated the testing at a later date due to a lack of usable stumble data, related to a combination a dropped trial, time used to determine an adequate perturbation length, and a leak of joint fluid that prevented completion of testing. Breaks were given between trials as needed.

Once laboratory testing was completed, participants were asked to complete a questionnaire assessing their level of their satisfaction with the OWS SCKAFO (Appendix H). Participants were reimbursed for parking expenses incurred from all sessions.

3.6.3 Measurements

Kinematic data were collected at 120 Hz to coincide with previous OWS testing on able-bodied participants (Lemaire 2012). All trials were also recorded on digital video. The research assistant manually recorded observations regarding participants' recovery responses to stumble events, false joint engagements (i.e. joint locked without provocation due to high walking speeds) and anecdotal participant feedback after each trial. Time elapsed, platform position, treadmill speed and distance, and split speed perturbation events were recorded in D-Flow for each trial.

3.6.4 Data analysis

Marker data were cropped using Vicon Nexus to include three TL gait strides preceding and three strides following stumble recovery before being exported to Visual 3D. Data were filtered with a dual pass 10Hz Butterworth filter and gait events were identified as described in Section 3.2.5. Lower body joint angles and angular velocities were extracted for TL, perturbation and recovery events and maximum and minimum values were extracted for TL, perturbation and recovery strides. Independent variables and dependent variables for kinematic evaluation were the same as those during stumble development. An additional independent variable was treadmill surface condition (TL, uphill, downhill, rolling hills, uneven). The dependent variable related to treadmill surface condition was knee collapse event (Yes or No). Differences in temporal spatial values, joint angles, and angular velocities for perturbed strides were compared with TL strides. TL stride values for right and left legs were not averaged, since differences between affected and unaffected legs showed differences between participants. Since all applied perturbations were a 'Right Trip', differences between left TL stride was compared to the perturbation stride values, and right

TL stride values were compared to recovery 1stride values. Manual data and digital videos were reviewed and treadmill speed outputs from D-Flow data were examined to determine changes in walking speed during perturbation recovery. Questionnaire results and anecdotal feedback observed during sessions were analyzed qualitatively to draw an understanding of user opinions toward the OWS.

Chapter 4. Results

4.1 Stumble development

4.1.1 Participants

Three male and two female participants, aged 22-32, completed testing in the RVR Lab. Table 6 shows participant information.

Table 6: Sex, age, weight, and height for able-bodied stumble development participants.

Participant	1	2	3	4	5
Sex	Male	Male	Female	Male	Female
Age (Years)	32	22	23	25	27
Weight (kg)	64.9	82.6	53.16	77.1	69.4
Height (cm)	181	183	161	188	170

4.1.2 Stumble recovery

The split speed perturbation invoked stumble outcomes in all trials for able-bodied participants (i.e., all participants had increased knee flexion at recovery 1 foot strike and during weight acceptance, compared to TL) (Appendix I). Knee flexion increased by 16.74° at REC1FS and 16.36° in weight acceptance for participants using the hopping strategy, while knee flexion increased 32.29° at REC1FS and 76.58° in weight acceptance for the participant using the stopping strategy.

Two different stumble recovery strategies were observed during this pilot test, hopping and stopping. Participants 1-4 used a hopping strategy, which involved quick steps following the perturbation to regain balance before resuming typical gait. The stopping strategy stalled forward movement following REC1FS. This strategy resulted in a crouching

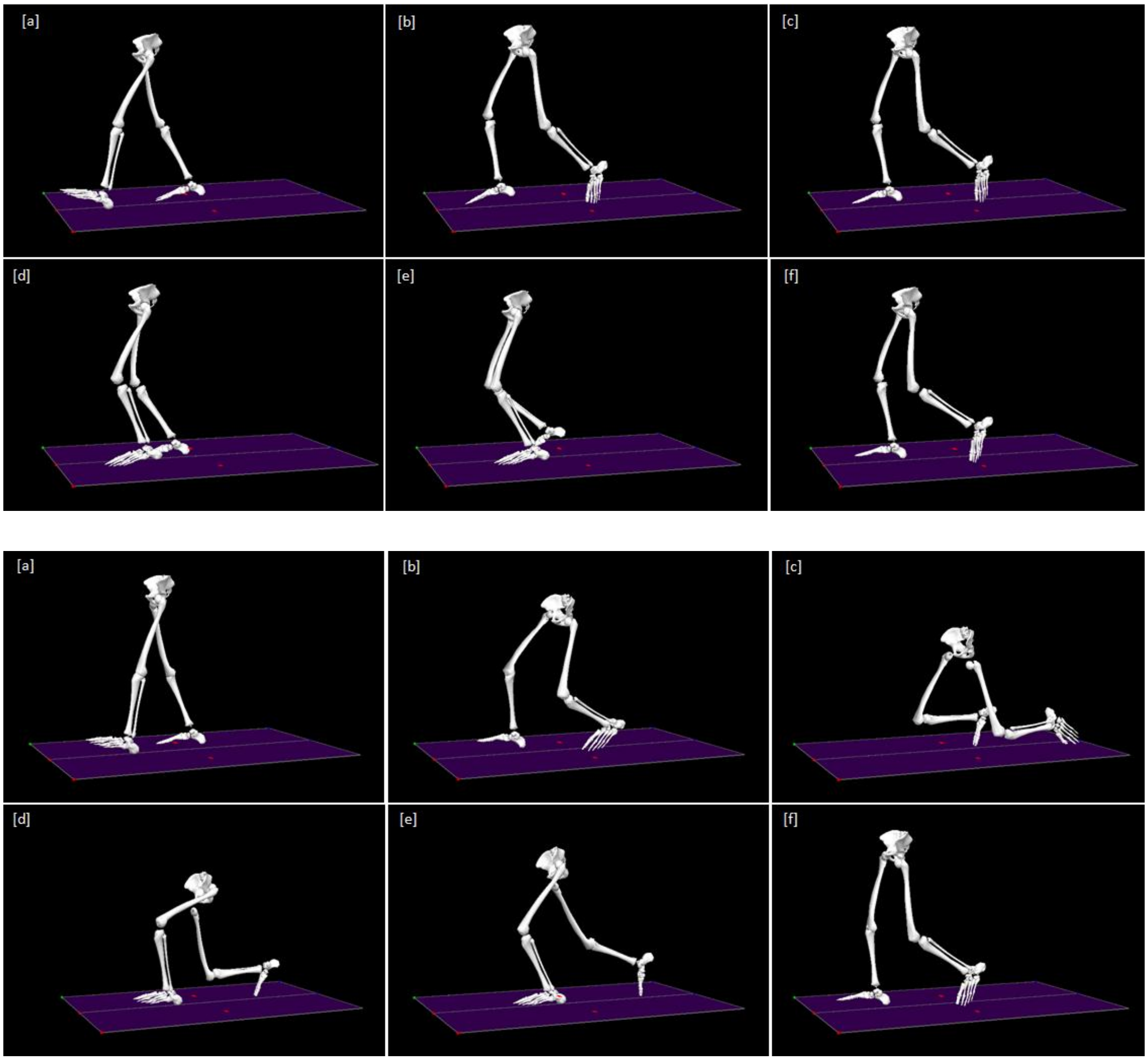


Figure 12: Visual 3D representation of lower body segments for hopping strategy (top) and stopping strategy (bottom) at a) PERTFS; b) REC1FS; c) PERTFO; d) REC2FS; e) REC1FO; and f) REC2FO.

action before regaining balance when the treadmill had slowed down, in some cases to a full stop. Handrail use was relied on heavily throughout the perturbation and for assistance pulling the body up out of the crouched position before continuing walking (Figure 12).

For participant 1, only 7 stumble trials were collected due to session time constraints. For participant 5, recovery 1 and recovery 2 step values for one trial were omitted due to an outlier trial, where the participant stopped and hanged in the harness before standing up and resuming gait. This counted as a fall, and recovery 1 and recovery 2 step values could not be defined. Kinematic results are presented separately for hopping and stopping strategies.

4.1.3 Temporal spatial parameters

Participants 1-4 used the hopping strategy, and increased their walking speed on average after the perturbation (1.52 ± 0.27 m/s) compared to before the perturbation (1.46 ± 0.28 m/s). On average, stance time decreased to 32% of TL during perturbation strides and 37% for recovery 1 strides; swing time decreased 21% for perturbation and trip strides; step length was similar to TL for perturbation strides (98%) and decreased to 85% of TL during recovery 1 strides; and DST decreased 30% compared to TL gait (Table 7).

Table 7: Temporal spatial parameters for hopping strategy.

	Stride	**Participant**				
		1	**2**	**3**	**4**	**5**
Stance Time (s)	TL	0.56 ± 0.02	0.71 ± 0.02	0.64 ± 0.02	0.62 ± 0.02	0.99 ± 0.43
	Recovery 1	0.43 ± 0.19	0.46 ± 0.05	0.33 ± 0.08	0.37 ± 0.10	2.18 ± 0.36
	Perturbation	0.41 ± 0.02	0.47 ± 0.02	0.42 ± 0.04	0.42 ± 0.03	1.46 ± 0.34
Swing Time (s)	TL	0.41 ± 0.01	0.39 ± 0.02	0.38 ± 0.01	0.41 ± 0.01	0.48 ± 0.27
	Recovery 1	0.33 ± 0.01	0.32 ± 0.02	0.30 ± 0.02	0.31 ± 0.02	0.27 ± 0.66
	Perturbation	0.29 ± 0.09	0.36 ± 0.03	0.34 ± 0.05	0.26 ± 0.04	0.53 ± 0.14
DST (s)	TL	0.16 ± 0.02	0.33 ± 0.03	0.25 ± 0.02	0.20 ± 0.02	0.49 ± 0.16
	Recovery 1	0.13 ± 0.13	0.15 ± 0.04	0.13 ± 0.06	0.14 ± 0.08	1.59 ± 0.47
Step Width (m)	TL	0.13 ± 0.05	0.15 ± 0.06	0.13 ± 0.05	0.16 ± 0.06	0.16 ± 0.08
	Recovery 1	0.11 ± 0.04	0.16 ± 0.02	0.06 ± 0.04	0.20 ± 0.06	0.19 ± 0.08
Step Length (m)	TL	0.81 ± 0.02	0.63 ± 0.03	0.64 ± 0.02	0.82 ± 0.03	0.52 ± 0.04
	Recovery 1	0.59 ± 0.08	0.61 ± 0.05	0.62 ± 0.04	0.59 ± 0.10	0.64 ± 0.21
	Perturbation	0.79 ± 0.04	0.62 ± 0.03	0.63 ± 0.02	0.81 ± 0.03	0.49 ± 0.06

Participant 5 used the stopping strategy and had the slowest walking speed leading up to the perturbation (0.79 ± 0.12 m/s) and reduced speed following the perturbation (0.45 ± 0.17 m/s), with an average minimum speed of 0.23 ± 0.17 m/s within 2s following perturbation. On average, stance times decreased to 45% of TL for perturbation stride and increased to 114% of TL for recovery 1 strides; swing time increased to 10% of TL for perturbation and decreased 44% for recovery 1 strides; step length was similar to TL for perturbations strides (94%) and increased to 123% of TL for recovery 1 strides; and DST increased 224% compared to TL gait.

Table 8: Hopping strategy joint angle averages and standard deviations (deg).

	TL FS	**PERTFS**	**REC1FS**	**REC2FS**
Ankle (X)	7.56 ± 1.86	6.81 ± 2.23	3.74 ± 7.77	11.41 ± 3.57
Knee (X)	7.87 ± 2.84	8.22 ± 2.75	24.52 ± 4.97	32.28 ± 5.20
Hip (X)	32.01 ± 8.70	31.96 ± 8.65	36.24 ± 6.76	29.06 ± 8.56
Pelvis (X)	7.72 ± 8.93	7.78 ± 8.95	13.05 ± 7.19	9.46 ± 7.53
Ankle (Y)	7.96 ± 3.00	7.64 ± 2.74	2.43 ± 1.50	6.85 ± 0.95
Knee (Y)	0.93 ± 3.14	0.81 ± 3.03	-1.86 ± 4.28	1.90 ± 3.45
Hip (Y)	-0.91 ± 3.41	-1.12 ± 3.06	0.74 ± 5.40	-2.99 ± 4.53
Pelvis (Y)	1.63 ± 1.53	1.60 ± 1.51	3.17 ± 0.52	-1.33 ± 0.52
	TL FO	**PERTFO**	**REC1FO**	**REC2FO**
Ankle (X)	-11.27 ± 3.54	-22.58 ± 4.13	2.83 ± 10.58	-8.31 ± 7.21
Knee (X)	47.72 ± 3.75	38.91 ± 5.95	48.16 ± 5.88	46.23 ± 3.69
Hip (X)	0.15 ± 6.74	0.99 ± 6.11	18.89 ± 4.90	6.97 ± 8.77
Pelvis (X)	5.97 ± 9.73	13.12 ± 7.33	10.32 ± 8.30	14.33 ± 10.29
Ankle (Y)	3.59 ± 3.75	4.53 ± 5.73	6.35 ± 7.85	4.57 ± 5.25
Knee (Y)	2.72 ± 2.65	2.43 ± 2.85	-0.26 ± 2.69	2.74 ± 1.97
Hip (Y)	-6.96 ± 2.59	-6.89 ± 3.79	-0.33 ± 4.72	-7.72 ± 4.23
Pelvis (Y)	-2.50 ± 1.51	-2.91 ± 1.06	1.37 ± 0.92	-2.41 ± 1.52

* (X): ankle dorsiflexion (+) and plantarflexion (-); knee flexion (+) and extension (-); hip flexion (+) and extension (-); pelvic anterior (+) and posterior (-) tilt. (Y): ankle inversion (+) and eversion (-); knee varus and valgus (-); hip adduction (+) and abduction (-); and pelvic upward (+) and downward (-) obliquity.

4.1.4 Hopping strategy kinematics

4.1.4.1 Perturbation stride

As the perturbation occurred, the treadmill belt of the perturbation limb sped up. All joint angles at PERTFS, as well as ankle plantar flexion at foot flat and knee flexion immediately following FS, were similar to TL gait (Table 8; Appendix J-M). Through stance, ankle velocity increased as the ankle dorsiflexed more quickly, compared to TL. In late stance, plantar flexion following FO was greater and appeared slightly earlier in the gait cycle than TL gait (Appendix N). Less knee flexion and hip extension, and greater pelvic anterior tilt occurred near PERTFO. At PERTFO, plantar flexion and pelvic tilt continued to be greater, while knee flexion was less than TL gait. All of these adjustments were made after the perturbation to quickly propel the leg back and off the sped up treadmill belt.

4.1.4.2 Recovery stride 1

At PERTFS, recovery 1 leg nears FO and prepares for swing. Following FO, ankle angles were less plantar flexed than TL. In mid to late swing following perturbation, ankle, knee and hip velocity peaks were greater than TL. This ensured fast ankle dorsiflexion, knee extension, and hip flexion to quickly bring the leg forward for REC1FS. Ankle dorsiflexion, hip flexion and pelvic anterior tilt peak angles increased during swing.

At REC1FS, hip extension velocity was greater than TL to rapidly move the leg behind the body through the shortened stance phase. Mildly less dorsiflexion and inversion were seen at the ankle due to a more flat footed landing compared to TL FS. Values for hip flexion, hip adduction, and pelvic anterior tilt were greater compared to TL FS. Knee flexion at REC1FS was considerably greater than TL (Figure 13), while knee varus alignment was slightly less.

In early stance, ankle angles for recovery 1 leg did not show a typical foot flat plantar flexion minimum. Two participants displayed plantar flexion earlier in the cycle, compared to TL, before or close to REC1FS. The other two people omitted this plantar flexion minimum with little change in dorsiflexion through stance. Knee angle values continued to be considerably greater than TL in early stance and knee angular velocity increased to extend the leg more quickly through the shortened stance phase. Knee and hip extension minimums, seen near TL FO, were less prominent near REC1FO. Peak pelvic anterior tilt during stance was greater than TL.

At REC1FO, ankle plantar flexion velocity was reduced as the ankle continued to be more dorsiflexed than TL for the shortened recovery steps. Hip flexion was relatively greater, ankle inversion and pelvic anterior tilt angles were mildly greater, and hip adduction was mildly reduced compared to TL.

4.1.4.3 Recovery 2 step

AT REC2FS, minimal ankle angular velocity indicated that the ankle was in a consistent state of dorsiflexion. Knee varus alignment was mildly greater compared to TL, while hip flexion, adduction, and pelvic obliquity were mildly lower. Knee flexion at REC2FS was considerably greater than TL. Hip extension velocity was increased to rapidly extend the hip under the body during stance.

At REC2FO, pelvic anterior tilt and hip flexion were moderately greater than TL. Ankle plantar flexion values returned to TL values, though ankle plantar flexion velocity was less compared to TL. Knee angles were similar to TL at REC2FS.

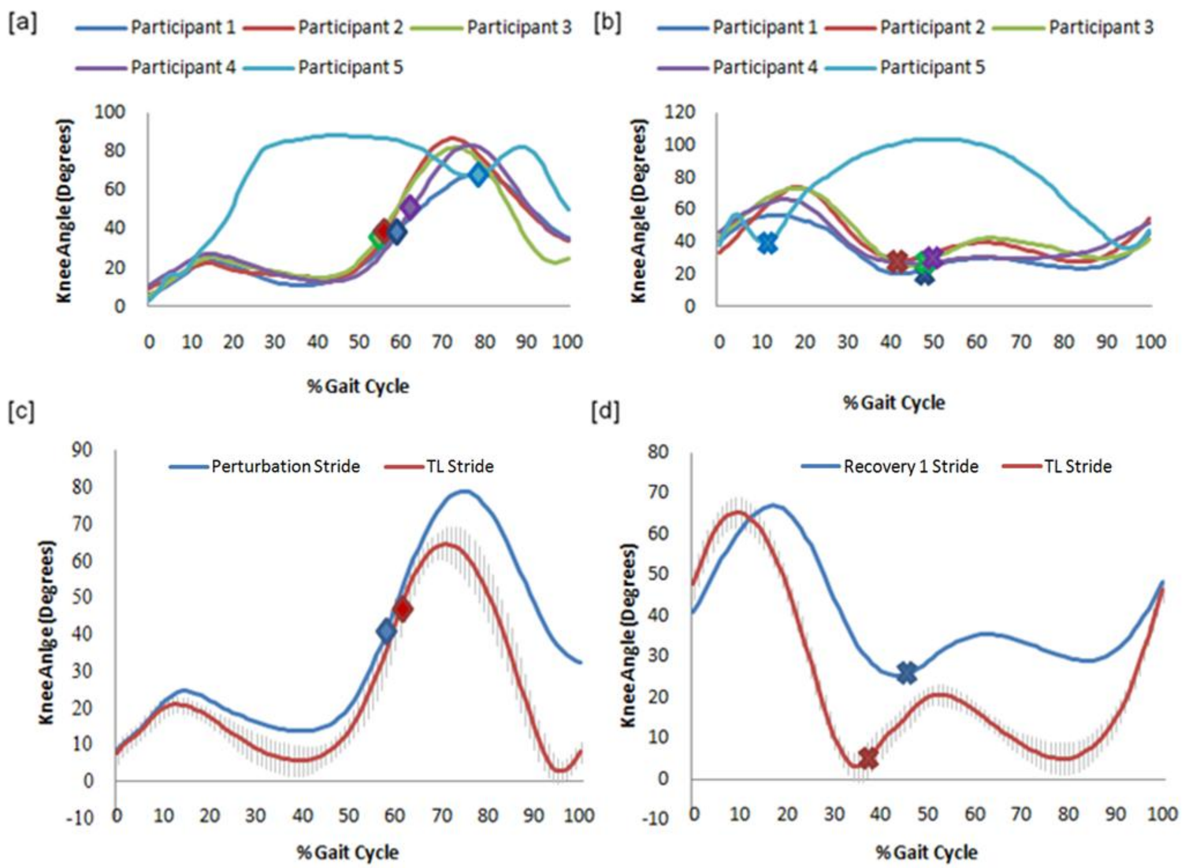


Figure 13: Knee flexon-extension (deg) normalized to 100% of the gait cycle.
[a] Perturbation stride for all participants. [b] Recovery 1 stride for all participants.
[c] Perturbation and TL strides for participants using hopping strategy. [d] Recovery 1 and TL strides for participants using hopping strategy. FO = ◇ ; FS = ✕.

4.1.5 Stopping strategy kinematics

4.1.5.1 Perturbation stride

At PERTFS, ankle, knee, hip, and pelvis angles were all similar to TL (Table 9). During stance, the perturbation leg moved rapidly backward with the increased treadmill speed, into a position that resembled kneeling to the ground. The ankle dorsiflexion angle following PERTFS was greater than TL, and then became variable between trials at mid stance. Knee flexion, hip flexion, and pelvic tilt were greater than TL at mid stance. The ankle was plantarflexed at PERTFO, but at TL FO the ankle was dorsiflexed. Knee flexion, hip flexion, and pelvic tilt were all greater than TL. Ankle, knee, and hip joint angular velocities were all closer to zero, indicating slower movement than at TL FO.

In swing of the perturbation stride, ankle dorsiflexion, knee and hip flexion, pelvic tilt were greater than TL. Ankle dorsiflexion velocity was greater and knee flexion velocity was less. Values during and following PERTFO were related to participant 5's initial actions to bring the body upward out of the crouched position to standing, using the safety handrails.

Table 9: Stopping strategy joint angle averages and standard deviations (deg).

	TL FS	**PERTFS**	**REC1FS**	**REC2FS**
Ankle (X)	7.97 ± 1.96	6.53 ± 2.25	7.71 ± 9.29	16.52 ± 6.09
Knee (X)	-0.11 ± 2.47	3.43 ± 3.35	32.18 ± 16.87	50.01 ± 39.28
Hip (X)	16.68 ± 3.12	16.56 ± 2.36	50.15 ± 11.59	62.84 ± 26.66
Pelvis (X)	3.13 ± 1.95	2.59 ± 1.14	16.11 ± 6.32	22.74 ± 7.64
Ankle (Y)	7.77 ± 2.68	5.70 ± 3.44	6.75 ± 3.36	4.13 ± 3.85
Knee (Y)	-2.16 ± 0.79	-2.08 ± 1.34	-0.37 ± 3.39	-4.07 ± 4.32
Hip (Y)	-1.05 ± 1.36	-1.34 ± 1.64	-6.60 ± 4.01	-4.39 ± 9.13
Pelvis (Y)	1.24 ± 0.82	0.58 ± 1.19	2.89 ± 3.81	2.48 ± 3.22
	TL FO	**PERTFO**	**REC1FO**	**REC2FO**
Ankle (X)	3.52 ± 4.44	-12.18 ± 19.60	-3.29 ± 14.12	1.10 ± 7.68
Knee (X)	46.84 ± 4.53	66.65 ± 26.39	47.41 ± 19.53	49.08 ± 10.35
Hip (X)	-0.62 ± 3.06	20.97 ± 25.90	8.77 ± 12.54	5.83 ± 7.78
Pelvis (X)	2.18 ± 2.34	29.58 ± 6.20	21.03 ± 12.13	8.25 ± 8.83
Ankle (Y)	11.44 ± 1.84	11.32 ± 9.15	8.82 ± 5.60	10.16 ± 2.51
Knee (Y)	-0.66 ± 3.28	0.40 ± 6.21	0.06 ± 2.40	0.82 ± 3.45
Hip (Y)	-9.52 ± 2.95	-15.98 ± 9.25	-10.57 ± 3.85	-12.20 ± 6.75
Pelvis (Y)	-3.36 ± 0.87	-5.68 ± 5.33	-4.61 ± 3.64	-4.83 ± 3.21

* (X): ankle dorsiflexion (+) and plantarflexion (-); knee flexion (+) and extension (-); hip flexion (+) and extension (-); pelvic anterior (+) and posterior (-) tilt. (Y): ankle inversion (+) and eversion (-); knee varus and valgus (-); hip adduction (+) and abduction (-); and pelvic upward (+) and downward (-) obliquity.

4.1.5.2 Recovery 1 stride

In swing, approaching REC1FS, ankle dorsiflexion, hip flexion, and pelvic tilt angles increased. At REC1FS, knee flexion, hip flexion, and pelvic anterior tilt increased, and the hip was slightly more abducted than TL. Ankle plantar flexion velocity and hip flexion velocity also increased.

In stance, forward movement was stopped, the treadmill slowed, and the person moved into a crouched position. As recovery 1 leg drew back, the ankle, knee, and hip absolute angular velocities decreased until the participant had slowed down enough to begin recovery. Ankle dorsiflexion, knee flexion, hip flexion and pelvic anterior tilt all increased after REC1FS over the lengthened stance phase, compared to TL. At REC1FO, as the participant brought the body upward out of the crouch with help from the handrails, the ankle plantar flexed and hip flexion and pelvic tilt were increased from TL. Ankle, knee, and hip joint velocities were all closer to zero due to the slowed speed.

4.1.5.3 Recovery 2 step

At REC2FS, ankle dorsiflexion, knee flexion, hip flexion, and pelvic tilt angles were all greater than TL. Ankle plantar flexion, knee flexion, and hip flexion velocities were also increased from TL as the participant continued to rise out of the crouched position.

At REC2FO, pelvic anterior tilt and hip flexion were still slightly greater than TL, while other angles were similar to TL FO values. Knee flexion and ankle plantar flexion were similar to TL, though ankle plantar flexion velocity decreased due to the slower walking speed.

4.2 OWS Testing

4.2.1 Participants

Four volunteers with knee extensor weakness agreed to participate in the OWS research. One volunteer was excluded from the study due to high knee extensor strength. The three male volunteers aged 51-59 years (Table 10). Participants provided informed consent before being fit with a custom OWS SCKAFO in the Prosthetics and Orthotics Service (P&O) at TOHRC, and completing OWS testing in the RVR Lab. This study was approved by The Ottawa Hospital and University of Ottawa Research Ethics Boards.

Table 10: MMT, orthosis history, and demographic information for OWS participants.

Participant	1	2	3
Age	51	55	59
Plantar Flexor MMT	Fused ankle	5	4
Dorsiflexor MMT	Fused ankle	5	4
Knee Flexor MMT	4	4	4-
Knee Extensor MMT	4	4	4-
Hip Flexor MMT	4+	5	4
Hip Extensor MMT	4+	5	5
Weight (kg)	124.5	87.1	80.7
Height (cm)	188	176	172
Braced Leg	Right	Right	Right
Years of orthosis use	3	1.5	2
Diagnosis	Traumatic nerve injury; right leg most affected. Fused ankle.	Traumatic nerve injury; right leg most affected. Fused ankle.	Surgical nerve injury; right leg most affected

4.2.2 Stumble recovery

OWS participants did not stumble during non-perturbed walking on any of the simulated surface conditions (TL, incline, decline, rolling hills, uneven). For this reason, analysis was restricted to each participant’s TL and perturbed gait.

The split speed perturbation invoked a stumble for OWS participants; however, the participants did not experience knee collapse upon weight-bearing. Though increased knee flexion angles were achieved in 4/5 trials for OWS participant 1, 5/5 trials for participant 2 and 3/5 trials for participant 3, participants quickly extended the knee through weight acceptance to less than normal or similar TL knee flexion angles (Table 11). Except for one trial, knee flexion max in early stance (Max 1: 0-25% of gait cycle) was equal to REC1FS. For all participants, the knee stayed very straight throughout the stance phase of recovery 1 stride. This recovery strategy is referred to as the straight-legged response. All participants used at least one handrail for support before, following, or during the perturbation.

Table 11: Knee flexion angles for TL FS, recovery 1 stride FS, maximum knee flexion angle from 0-25% of the gait cycle (Max1), and knee flexion angle at 12% of the gait cycle for TL and recovery 1 stride.

	Trial	TL FS (Right)	REC1FS	Max 1	TL at 12%	Recovery 1 at 12%
Participant 1	1	2.89	10.05	10.05	1.87	-10.55
	2	4.45	16.16	16.16	1.04	0.98
	3	3.29	15.42	15.42	3.66	-8.57
	4	3.18	15.81	15.81	3.41	-9.94
	5	5.48	3.74	3.74	4.83	-2.90
Participant 2	1	23.97	31.12	31.12	31.32	3.41
	2	28.11	37.97	37.97	33.19	3.41
	3	16.89	17.03	35.32	21.34	4.01
	4	21.40	43.35	43.42	27.99	3.42
	5	24.20	34.51	34.51	32.45	3.51
Participant 3	1	3.07	4.72	4.72	-6.49	-6.42
	2	1.24	10.87	10.87	-6.57	-6.62
	3	1.46	11.06	11.06	-6.75	-6.21
	4	1.26	-0.76	-0.76	-6.63	-6.56
	5	3.13	1.42	1.42	-6.72	-6.85

Though the straight-legged strategy was used for recovery after all perturbations, three trials stood out as different. Participant 2 quickly aborted the perturbation stride in two trials by shifting to the braced leg to regain stability. This resulted in a negative stride length for these two trails. Participant 3, during the final stumble trial, managed to avoid the treadmill belt perturbation. This was achieved by remaining in stance on the braced leg and not putting the unaffected leg down until the treadmill perturbation had completed.

All kinematic results are presented individually.

4.2.3 Temporal spatial parameters

4.2.3.1 Participant 1

Over the complete perturbation cycle, participant 1 slowed down from 0.79 ± 0.08 m/s in the 2 seconds before the perturbation to 0.68 ± 0.13 m/s in the 2 seconds after the perturbation. The average minimum speed following the perturbation was 0.56 ± 0.14 m/s, with a range of 0.32 m/s to 0.68 m/s.

Following the perturbation, participant 1 increased DST to 135% of TL. In the perturbation stride, step width, step length and stance time were decreased to 94%, 90% and 77% of TL, respectively, and swing time was increased to 181% of TL. In recovery 1 stride, swing time leading to REC1FS was decreased to 48% of TL and step width, step length and stance time were increased to 123%, 134%, and 170% of TL. Table 12 presents temporal spatial parameters for TL, perturbation, and recovery 1 strides.

Table 12: Temporal spatial parameters for OWS participants. Perturbation refers to the left (L) non-braced leg; Recovery 1 refers to the right (R) braced leg.

	Participant 1			
Temporal Spatial	**TL (L)**	**Perturbation (L)**	**TL (R)**	**Recovery 1 (R)**
Stance Time (s)	1.07 ± 0.07	0.83 ± 0.05	1.08 ± 0.07	1.84 ± 0.49
Swing Time (s)	0.54 ± 0.03	0.98 ± 0.34	0.54 ± 0.04	0.26 ± 0.03
Stride Time (s)	1.62 ± 0.10	1.80 ± 0.34	1.62 ± 0.10	2.10 ± 0.48
Step Length (m)	0.59 ± 0.05	0.53 ± 0.06	0.65 ± 0.06	0.87 ± 0.08
Step Width (m)	0.17 ± 0.01	0.16 ± 0.01	0.17 ± 0.01	0.21 ± 0.03
DST (s)	0.53 ± 0.05	0.72 ± 0.55	0.53 ± 0.05	0.72 ± 0.55
	Participant 2			
Temporal Spatial	**TL (L)**	**Perturbation (L)**	**TL (R)**	**Recovery 1 (R)**
Stance Time (s)	0.92 ± 0.10	0.63 ± 0.16	0.93 ± 0.10	1.21 ± 0.28
Swing Time (s)	0.43 ± 0.03	0.67 ± 0.16	0.42 ± 0.03	0.10 ± 0.09
Stride Time (s)	1.35 ± 0.13	1.30 ± 0.28	1.35 ± 0.13	1.31 ± 0.34
Step Length (m)	0.52 ± 0.07	0.50 ± 0.07	0.51 ± 0.05	0.28 ± 0.28
Step Width (m)	0.16 ± 0.01	0.16 ± 0.00	0.16 ± 0.01	0.17 ± 0.01
DST (s)	0.51 ± 0.09	0.54 ± 0.19	0.51 ± 0.09	0.54 ± 0.19
	Participant 3			
Temporal Spatial	**TL (L)**	**Perturbation (L)**	**TL (R)**	**Recovery 1 (R)**
Stance Time (s)	1.64 ± 0.06	1.38 ± 0.34	1.44 ± 0.04	1.41 ± 0.18
Swing Time (s)	0.54 ± 0.03	0.70 ± 0.14	0.73 ± 0.04	0.59 ± 0.16
Stride Time (s)	2.16 ± 0.08	2.08 ± 0.32	2.17 ± 0.07	2.00 ± 0.13
Step Length (m)	0.46 ± 0.03	0.37 ± 0.06	0.43 ± 0.02	0.44 ± 0.03
Step Width (m)	0.13 ± 0.01	0.12 ± 0.02	0.12 ± 0.01	0.15 ± 0.08
DST (s)	0.91 ± 0.06	0.78 ± 0.28	0.91 ± 0.06	0.78 ± 0.28

4.2.3.2 Participant 2

Participant 2 slowed from 0.81 ±0.13 m/s before the perturbation to 0.65 ± 0.26 m/s after the perturbation. The average minimum speed following the perturbation was 0.56 m/s, ranging from 0.30m/s to 0.85m/s.

Following the perturbation, participant 2 increased DST to 106% of TL. In the perturbation stride, step width was equal compared to TL, step length and stance time decreased to 96% and 68% of TL, respectively, and swing time increased to 156% of TL.

In recovery 1 stride, swing time and step length decreased to 24% and 55% of TL, and step width and stance time increased to 106% and 130% of TL, respectively.

4.2.3.3 Participant 3

Participant 3 slowed from 0.39 ± 0.03 m/s before the perturbation to 0.26 ± 0.14 m/s following the perturbation. The average minimum speed following the perturbation was 0.19 ± 0.15 m/s, with a range of 0.0 m/s to 0.35m/s.

Unlike participants 1 and 2, participant 3 decreased DST to 86% of TL following the perturbation. In the perturbation stride, step width, step length, and stance time decreased to 92%, 80% and 84% of TL, respectively, and swing time increased to 130% of TL. Step width increased in recovery 1 stride to 125% of TL, and unlike participants 1 and 2, step length, stance time, and swing time were similar to TL at 102%, 98% and 96%, respectively.

4.2.4 Kinematics: Participant 1

4.2.4.1 Perturbation stride

At PERTFS, all joint angles were similar to TL FS (Table 13; Appendix O-Q). Ankle plantar flexion velocity was greater upon PERTFS due to the increased treadmill belt speed upon foot contact. In stance, ankle dorsiflexion was less than TL. Knee flexion and hip flexion angles were both slightly greater through stance, though knee extension and hip extension velocities were also greater (Appendix R). These values coincided with a quick backward movement of the perturbation leg on the treadmill, while the participant leaned forward at the trunk and used the safety handrails for support (Figure 14).

Table 13: Average and standard deviation for angles (degrees) at TL FS and FO and perturbed FS and FO events for OWS Participant 1.

	Plane	Non-Braced			Braced	
		Left TL FS	**PERTFS**	**REC2FS**	**Right TL FS**	**REC1FS**
Ankle	X	4.81 ± 0.54	3.83 ± 1.13	0.34 ± 1.44	8.41 ± 0.67	10.99 ± 1.22
	Y	1.12 ± 0.43	0.39 ± 0.56	-0.58 ± 0.66	6.46 ± 0.19	6.26 ± 0.36
	Z	1.71 ± 0.70	2.04 ± 1.23	-2.80 ± 1.62	13.25 ± 0.16	12.06 ± 0.60
Knee	X	5.83 ± 1.55	7.28 ± 4.55	28.59 ± 6.57	3.86 ± 1.08	12.23 ± 5.37
	Y	-2.63 ± 0.61	-2.18 ± 1.88	6.62 ± 3.48	-3.84 ± 0.08	-3.87 ± 0.22
	Z	-20.19 ± 3.16	-19.44 ± 3.64	-11.60 ± 5.29	-5.58 ± 0.17	-6.16 ± 0.65
Hip	X	47.77 ± 4.08	47.03 ± 3.17	68.50 ± 9.26	48.90 ± 5.03	67.49 ± 3.83
	Y	5.17 ± 2.19	4.63 ± 2.89	0.89 ± 4.43	2.01 ± 1.28	7.01 ± 2.52
	Z	2.52 ± 2.06	1.55 ± 3.72	11.22 ± 4.08	-16.93 ± 2.11	-26.66 ± 3.61
Pelvis	X	23.18 ± 2.80	22.62 ± 1.54	40.06 ± 6.88	23.28 ± 3.53	37.58 ± 2.20
	Y	-4.22 ± 2.44	-3.35 ± 2.22	-0.87 ± 5.52	-1.10 ± 2.19	15.51 ± 3.41
	Z	3.26 ± 1.67	4.30 ± 2.12	-1.53 ± 2.01	2.90 ± 1.68	3.98 ± 2.95
	Plane	**Non-Braced**			**Braced**	
		Left TL FO	**PERTFO**	**REC2FO**	**Right TL FO**	**REC1FO**
Ankle	X	4.36 ± 0.97	-5.91 ± 4.22	2.18 ± 3.06	7.50 ± 2.42	12.04 ± 6.96
	Y	1.34 ± 0.61	-0.36 ± 1.07	0.67 ± 1.35	6.14 ± 0.57	7.76 ± 1.61
	Z	1.90 ± 0.56	-0.33 ± 4.49	-1.26 ± 2.00	10.16 ± 0.85	9.78 ± 0.99
Knee	X	43.75 ± 1.92	38.20 ± 3.99	46.37 ± 7.89	50.67 ± 3.41	35.58 ± 17.42
	Y	10.49 ± 0.66	7.29 ± 2.59	14.25 ± 3.76	-3.54 ± 0.22	-3.59 ± 0.17
	Z	-5.87 ± 1.47	-5.38 ± 3.22	-8.66 ± 1.38	-9.18 ± 0.26	-7.73 ± 1.38
Hip	X	22.01 ± 1.76	20.67 ± 3.74	37.34 ± 10.34	17.30 ± 2.79	22.27 ± 11.07
	Y	-9.19 ± 1.16	-20.86 ± 2.40	-11.67 ± 4.98	-11.61 ± 1.69	-8.40 ± 5.87
	Z	-7.06 ± 3.05	-5.47 ± 3.57	2.49 ± 8.96	-19.11 ± 1.48	-22.14 ± 5.12
Pelvis	X	25.84 ± 4.30	44.18 ± 4.63	39.28 ± 3.46	20.01 ± 3.39	37.62 ± 4.77
	Y	1.68 ± 1.99	17.22 ± 1.42	3.91 ± 5.06	-5.34 ± 2.35	-1.86 ± 5.41
	Z	1.97 ± 0.80	3.21 ± 3.82	4.16 ± 2.22	0.25 ± 3.21	-2.04 ± 2.97

* Ankle: X = dorsiflexion (+) and plantarflexion (-); Y = inversion (+) and eversion (-); Z = adduction (+) and abduction (-). Knee: X = flexion (+) and extension (-); Y = varus (+) and valgus (-); Z = internal (+) and external (-) rotation. Hip: X = flexion (+) and extension (-); Y = adduction (+) and abduction (-); Z = internal (+) and external (-) rotation. Pelvis: X = anterior (+) and posterior (-) tilt; Y = pelvic upward (+) and downward (-) obliquity; Z = internal (+) and external (-) rotation.

Around PERTFO, ankle dorsiflexion velocity was variable and greater than TL, while knee flexion velocity was highly variable and less than TL. This variability was due to the participant's tendency to drag the toe forward on the treadmill after PERTFO in the initial portion of swing phase.

At PERTFO, ankle plantar flexion and slight ankle eversion and abduction contrasted with ankle dorsiflexion and slight ankle inversion and adduction values that occurred at TL FO. Flexion and varus alignment at the knee were slightly less at PERTFO, while hip adduction was considerably less and pelvic tilt and obliquity were considerably greater than TL. Ankle plantar flexion, knee flexion, and hip flexion angular velocities all decreased as the treadmill slowed. Knee flexion in swing before REC2FS was greater than TL.

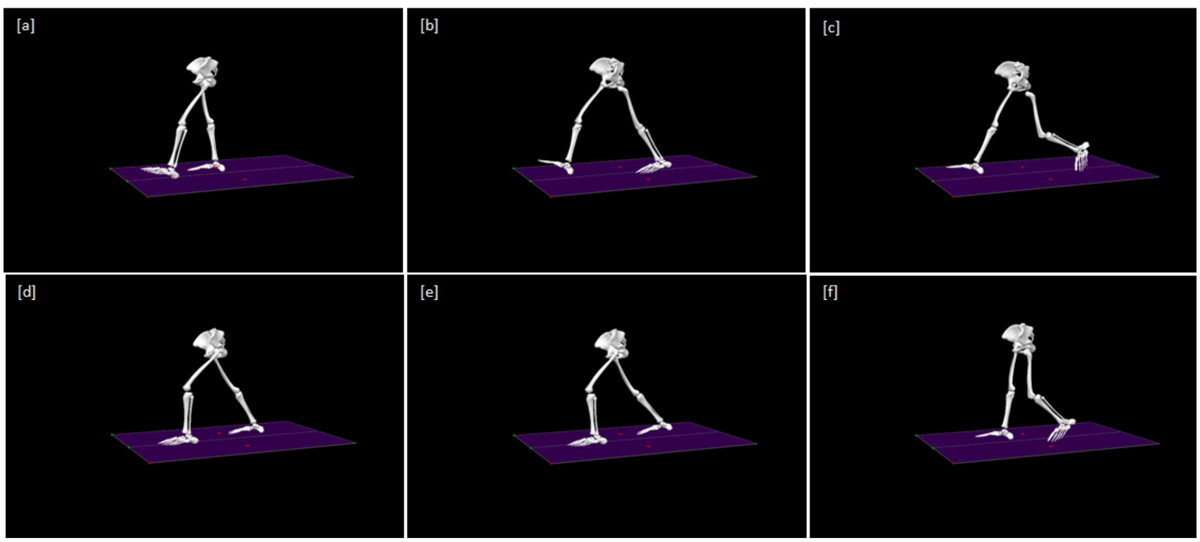


Figure 14: Visual 3D representation of lower body segments for OWS Participant 1 at a) PERTFS; b) REC1FS; c) PERTFO; d) REC2FS; e) REC1FO; and f) REC2FO.

4.2.4.2 Recovery 1 stride

Hip flexion velocity was greater than TL in swing, while the participant quickly brought the leg forward to prepare for REC1FS. Knee flexion in swing was slightly less than TL and the knee extension velocity was greater, but at REC1FS the knee flexion angle was moderately greater compared to TL. This is because, following the stumble, the flexion angle seen in late swing during TL gait occurs near REC1FS, likely due to the reduced swing time allowed to bring the braced leg back into the typical extended position for weight acceptance. However, in early stance, the knee hyperextended (i.e, typically flexed), and knee extension angular velocity was greater than TL. These results corresponded with the participant's tendency to quickly straighten the leg through weight acceptance, with the assistance of the handrails, as a strategy to prevent knee collapse (Figure 15).

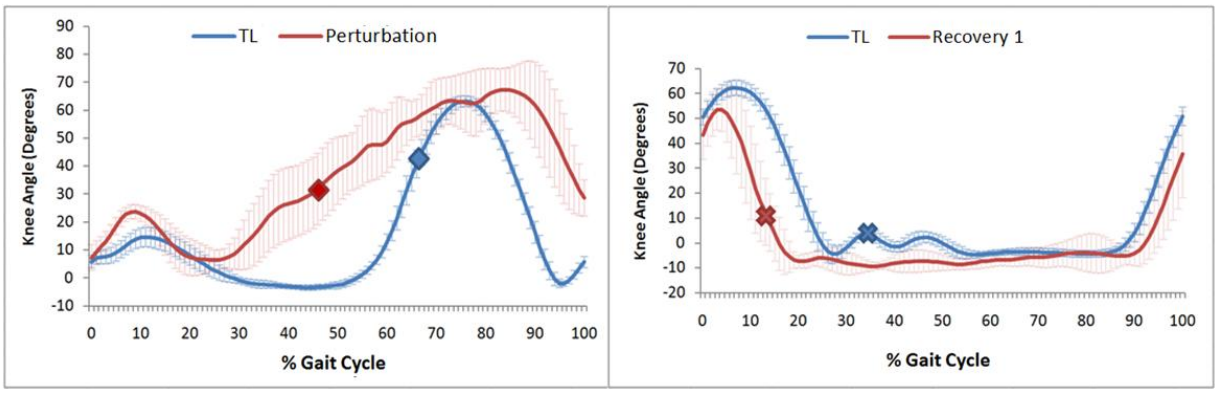


Figure 15: Knee flexion (+) and extension (-) angles for OWS Participant 1: from FS to FS for the left leg for TL and perturbation strides (left), and from FO to FO for the right leg for TL and recovery 1 strides (right).

Ankle dorsiflexion angle at REC1FS was slightly greater compared to TL, while there was considerably more hip flexion and anterior pelvic tilt. The hip was more adducted and externally rotated at REC1FS, and positive pelvic obliquity was greater compared to TL. In early stance, ankle plantar flexion angle and velocity were greater than TL through foot flat, and hip flexion and pelvic anterior tilt were also greater. Ankle dorsiflexion velocity was less than TL in mid stance.

Ankle dorsiflexion and hip flexion were slightly greater at REC1FO, and the hip was slightly more abducted and externally rotated. Negative pelvic obliquity was slightly less, pelvic rotation was negative, and knee flexion was less compared to TL. Ankle plantar flexion velocity was less than at TL FO.

4.2.4.3 Recovery 2 step

At REC2FS, ankle dorisflexion, hip adduction, and negative pelvic obliquity were slightly less than TL. Compared to TL FS, the ankle was abducted (i.e., typically adducted for TL), and the knee was internally rotated (i.e., typically externally rotated). Knee flexion, hip flexion, hip internal rotation, and pelvic anterior tilt were considerably greater than TL at REC2FS, and knee extension velocity contrasted with knee flexion velocity seen at TL FS.

At REC2FO, the ankle was slightly less dorsiflexed than TL. The ankle was also abducted, which contrasted with the ankle adduction seen at FO in TL gait. Slightly greater angles were seen for knee flexion, varus alignment and external rotation, hip abduction, pelvic obliquity, and pelvic internal rotation. Hip flexion and pelvic anterior tilt angles were considerably greater compared to TL, while hip internal rotation values contrasted with external rotation seen at TL FO.

4.2.5 Kinematics: Participant 2

4.2.5.1 Perturbation stride

At PERTFS, all angles were similar to TL FS while knee flexion velocity was greater than TL (Table 14). Following PERTFS, in early stance, knee flexion angle was greater than TL, and ankle dorsiflexion and hip extension velocities were greater as the perturbed leg drew back quickly on the treadmill. Ankle plantar flexion approaching FO was greater, and appeared slightly earlier in the gait cycle than TL.

Table 14: Average and standard deviation for angles (degrees) at TL FS and FO and perturbed FS and FO events for OWS Participant 2.

	Plane	Non-Braced			Braced	
		TL Left FS	PERTFS	REC2FS	TL Right FS	REC1FS
Ankle	X	1.71 ± 3.33	0.59 ± 2.88	2.84 ± 5.46	8.80 ± 2.08	10.97 ± 6.96
	Y	5.08 ± 0.38	3.70 ± 0.93	6.16 ± 1.75	6.36 ± 0.22	6.76 ± 1.00
	Z	16.01 ± 1.08	15.72 ± 1.30	14.76 ± 1.58	10.83 ± 0.43	11.49 ± 0.44
Knee	X	13.71 ± 3.89	14.01 ± 3.58	14.82 ± 2.93	22.91 ± 4.14	32.80 ± 9.90
	Y	-6.18 ± 0.59	-6.20 ± 0.59	-5.61 ± 0.41	-3.63 ± 0.23	-3.60 ± 0.45
	Z	-10.76 ± 0.92	-10.15 ± 1.74	-7.49 ± 1.34	-10.61 ± 0.37	-11.76 ± 1.46
Hip	X	62.22 ± 5.17	61.62 ± 4.57	60.88 ± 2.93	62.56 ± 4.53	53.49 ± 10.37
	Y	0.25 ± 2.12	0.13 ± 1.48	0.25 ± 3.90	2.18 ± 0.82	5.42 ± 2.04
	Z	-7.74 ± 2.59	-8.39 ± 2.23	-8.29 ± 2.19	-15.14 ± 1.21	-9.97 ± 2.47
Pelvis	X	30.91 ± 2.81	31.07 ± 2.89	32.43 ± 1.88	29.48 ± 2.03	34.09 ± 1.09
	Y	3.93 ± 1.89	4.17 ± 1.91	3.10 ± 2.87	4.72 ± 0.63	4.92 ± 1.66
	Z	2.08 ± 1.45	2.33 ± 0.74	2.29 ± 2.81	-2.23 ± 1.57	0.74 ± 1.41
	Plane	**Non-Braced**			**Braced**	
		TL Left FO	**PERTFO**	**REC2FO**	**TL Right FO**	**REC1FO**
Ankle	X	-7.94 ± 3.39	-2.92 ± 7.48	-6.72 ± 4.18	4.84 ± 1.23	6.96 ± 2.64
	Y	5.47 ± 0.56	4.36 ± 1.52	6.87 ± 1.67	6.16 ± 0.35	6.57 ± 0.69
	Z	11.42 ± 1.73	11.05 ± 2.94	11.63 ± 3.39	11.20 ± 0.27	11.23 ± 0.26
Knee	X	56.66 ± 2.16	41.47 ± 4.73	55.83 ± 4.41	66.94 ± 0.79	66.04 ± 4.27
	Y	-3.51 ± 0.53	-3.69 ± 1.37	-3.34 ± 0.85	-2.47 ± 0.10	-2.70 ± 0.29
	Z	-5.72 ± 1.40	-6.99 ± 1.53	-5.38 ± 0.94	-15.77 ± 0.04	-15.82 ± 0.52
Hip	X	41.35 ± 2.37	33.50 ± 9.08	41.41 ± 3.34	40.29 ± 0.73	43.53 ± 4.78
	Y	-6.82 ± 1.95	-3.43 ± 2.47	-6.20 ± 1.88	1.92 ± 0.41	2.07 ± 1.59
	Z	-8.34 ± 0.79	-11.08 ± 1.11	-9.53 ± 1.95	-9.86 ± 0.52	-7.74 ± 0.81
Pelvis	X	29.90 ± 2.66	37.81 ± 3.03	31.21 ± 1.76	29.37 ± 2.80	30.27 ± 2.32
	Y	4.35 ± 1.26	4.83 ± 1.97	3.37 ± 1.98	3.82 ± 0.63	3.07 ± 1.00
	Z	-0.45 ± 1.48	-2.89 ± 4.46	1.49 ± 3.26	-0.73 ± 1.88	-1.46 ± 3.57

[*] Ankle: X = dorsiflexion (+) and plantarflexion (-); Y = inversion (+) and eversion (-); Z = adduction (+) and abduction (-). Knee: X = flexion (+) and extension (-); Y = varus (+) and valgus (-); Z = internal (+) and external (-) rotation. Hip: X = flexion (+) and extension (-); Y = adduction (+) and abduction (-); Z = internal (+) and external (-) rotation. Pelvis: X = anterior (+) and posterior (-) tilt; Y = pelvic upward (+) and downward (-) obliquity; Z = internal (+) and external (-) rotation.

At PERTFO, hip flexion and hip adduction were less, while hip external rotation and pelvic external rotation were slightly greater than TL. Ankle dorsiflexion velocity at PERTTO contrasted with plantar flexion velocity seen at TL FO, and ankle plantar flexion was less than TL as a result. Pelvic anterior tilt around PERTFO was greater, while knee flexion at PERTFO was considerably less than TL. Knee flexion angle continued to be slightly less in swing, as the participant used the handlebars for support and brought the leg forward using the hip (Figure 16).

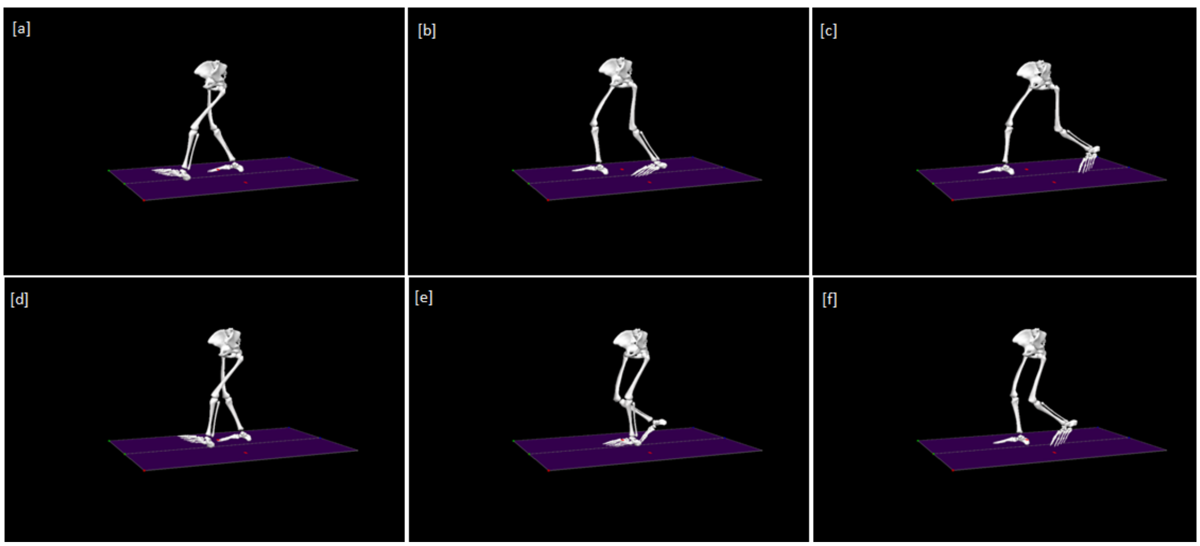


Figure 16: Visual 3D representation of lower body segments for OWS Participant 2 at a) PERTFS; b) REC1FS; c) PERTFO; d) REC2FS; e) REC1FO; and f) REC2FO.

4.2.5.2 Recovery 1 stride

Leading up to REC1FS, knee extension velocity was greater than TL and knee flexion angle was considerably less in the braced leg. At REC1FS ankle dorsiflexion, hip adduction and pelvic tilt were all slightly greater compared to TL FS. Hip external rotation was slightly less, and slight positive rotation of the pelvis contrasted with negative rotation typically seen at FS in TL gait. Knee extension and hip flexion velocities at REC1FS contrasted with knee flexion and hip extension velocities seen in TL gait, which corresponds with the participant quickly straightening the leg to bring it under the body for FS. Due to the shortened swing time, hip flexion angle was less and knee flexion was greater than TL.

Similar to Participant 1, the knee extension velocity was greater following REC1FS, as the participant quickly straightened the knee (Figure 17). Knee angles returned to typical TL values through weight acceptance and were less than TL in mid stance, likely due to a preventative strategy to avoid the threat of knee collapse during instability. Ankle dorsiflexion is also less during the stance phase, while hip flexion and pelvic tilt are greater than TL. Joint angles at REC1FO are mostly similar to TL; ankle dorsiflexion and hip flexion were slightly increased while hip external rotation was slightly less. All joint angular velocities were similar to TL at REC1FO.

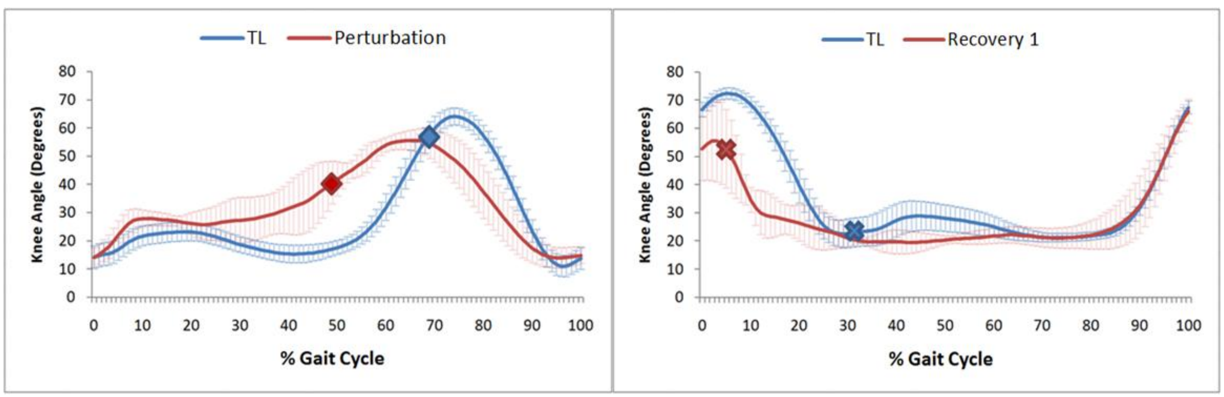


Figure 17: Knee flexion (+) and extension (-) angles for OWS Participant 2: from FS to FS for the left leg for TL and perturbation strides (left), and from FO to FO for the right leg for TL and recovery 1 strides (right).

4.2.5.3 Recovery 2 step

Joint angle values at REC2FS are all similar to TL, with the exception of knee external rotation, which is slightly less than TL. All joint angle values at REC2FO are similar to those at TL FO. All joint angular velocities were similar to TL at REC2FS and REC2FO.

4.2.6 Kinematics: Participant 3

4.2.6.1 Perturbation stride

Participant 3 was the only person to display slight differences in joint angles at PERTFS (Table 15). Ankle dorsiflexion and knee flexion were slightly greater than TL, while angles were slightly less for knee external rotation, hip flexion, and pelvic tilt. Knee flexion velocity was less than TL and hip extension velocity was slightly greater. Knee flexion angle following PERTFS was slightly greater and continued to be greater through mid-stance, as the perturbed leg moved quickly backward on the treadmill belt. Pelvic tilt was greater in mid stance and ankle dorsiflexion was less in late stance (Figure 18).

At PERTFO the ankle was slightly more adducted than TL and hip flexion and pelvic tilt were greater. Ankle plantar flexion velocity at PERTFO contrasted with dorsiflexion velocity seen at TL FO, and knee extension velocity was greater compared to TL, resulting in a smaller knee flexion angle at PERTFO. Slight varus alignment at the knee contrasted with valgus alignment seen at TL FO. In swing, ankle dorsiflexion and knee extension velocities were greater compared to TL.

Table 15: Average and standard deviation for angles (degrees) at TL FS and FO and perturbed FS and FO events for OWS Participant 3.

	Plane	Non-Braced			Braced	
		Left FS	**PERTFS**	**REC2FS**	**Right FS**	**REC1FS**
Ankle	X	2.21 ± 2.10	4.18 ± 3.62	4.99 ± 8.59	-3.76 ± 0.85	-1.44 ± 4.82
	Y	2.09 ± 1.25	2.46 ± 1.81	2.50 ± 2.60	-1.22 ± 0.28	-1.78 ± 0.33
	Z	12.13 ± 0.55	10.44 ± 1.40	11.08 ± 1.52	12.33 ± 0.46	12.37 ± 0.35
Knee	X	31.48 ± 3.01	37.27 ± 3.23	37.43 ± 18.44	2.03 ± 0.98	5.46 ± 5.39
	Y	2.06 ± 0.78	3.03 ± 1.59	3.10 ± 1.63	9.56 ± 0.16	9.31 ± 0.66
	Z	-14.42 ± 1.53	-11.94 ± 3.67	-13.21 ± 2.31	-15.73 ± 0.22	-15.73 ± 0.42
Hip	X	46.39 ± 0.60	44.33 ± 2.34	46.70 ± 6.01	23.33 ± 1.81	30.69 ± 8.37
	Y	-1.98 ± 1.24	-3.76 ± 0.80	-3.08 ± 3.92	-0.99 ± 1.43	-4.94 ± 3.48
	Z	-1.40 ± 0.80	-1.89 ± 0.50	-1.58 ± 2.01	-12.60 ± 2.74	-16.51 ± 2.86
Pelvis	X	9.39 ± 1.27	5.87 ± 1.56	11.00 ± 5.89	9.24 ± 1.87	20.83 ± 6.49
	Y	5.05 ± 0.93	6.08 ± 1.02	4.92 ± 2.03	6.43 ± 1.30	9.63 ± 1.83
	Z	-3.99 ± 2.36	-5.00 ± 1.74	2.33 ± 10.70	-3.09 ± 1.48	-7.06 ± 4.99
	Plane	**Non-Braced**			**Braced**	
		Left FO	**PERTTO**	**REC2FO**	**Right FO**	**REC1FO**
Ankle	X	-6.02 ± 1.75	-4.28 ± 6.54	-1.66 ± 3.27	-2.01 ± 0.51	-1.43 ± 1.82
	Y	1.72 ± 1.16	3.10 ± 2.09	2.78 ± 2.74	1.00 ± 0.15	0.17 ± 0.84
	Z	11.11 ± 1.14	13.13 ± 1.06	13.58 ± 2.09	12.54 ± 0.97	12.66 ± 2.23
Knee	X	74.55 ± 1.77	66.43 ± 11.17	74.59 ± 2.14	24.49 ± 2.56	14.10 ± 13.23
	Y	-2.60 ± 1.52	0.14 ± 1.03	-2.31 ± 3.46	7.20 ± 0.35	8.34 ± 1.49
	Z	-7.71 ± 0.61	-6.59 ± 0.63	-7.35 ± 1.48	-14.63 ± 0.20	-15.27 ± 0.55
Hip	X	24.28 ± 2.36	27.25 ± 5.20	29.17 ± 6.25	4.97 ± 0.91	7.08 ± 5.82
	Y	-10.44 ± 1.42	-11.02 ± 1.04	-9.55 ± 3.73	0.30 ± 0.96	-1.43 ± 3.13
	Z	-3.93 ± 0.51	-4.96 ± 2.19	-2.80 ± 3.52	-6.80 ± 2.08	-10.65 ± 10.35
Pelvis	X	6.40 ± 2.36	16.79 ± 7.08	7.21 ± 1.33	17.04 ± 1.59	15.58 ± 6.38
	Y	6.28 ± 1.29	6.98 ± 1.43	6.41 ± 0.98	6.56 ± 1.35	7.01 ± 1.66
	Z	-8.72 ± 2.61	-9.97 ± 4.47	-14.00 ± 10.51	5.98 ± 1.93	13.53 ± 10.45

* Ankle: X = dorsiflexion (+) and plantarflexion (-); Y = inversion (+) and eversion (-); Z = adduction (+) and abduction (-). Knee: X = flexion (+) and extension (-); Y = varus (+) and valgus (-); Z = internal (+) and external (-) rotation. Hip: X = flexion (+) and extension (-); Y = adduction (+) and abduction (-); Z = internal (+) and external (-) rotation. Pelvis: X = anterior (+) and posterior (-) tilt; Y = pelvic upward (+) and downward (-) obliquity; Z = internal (+) and external (-) rotation.

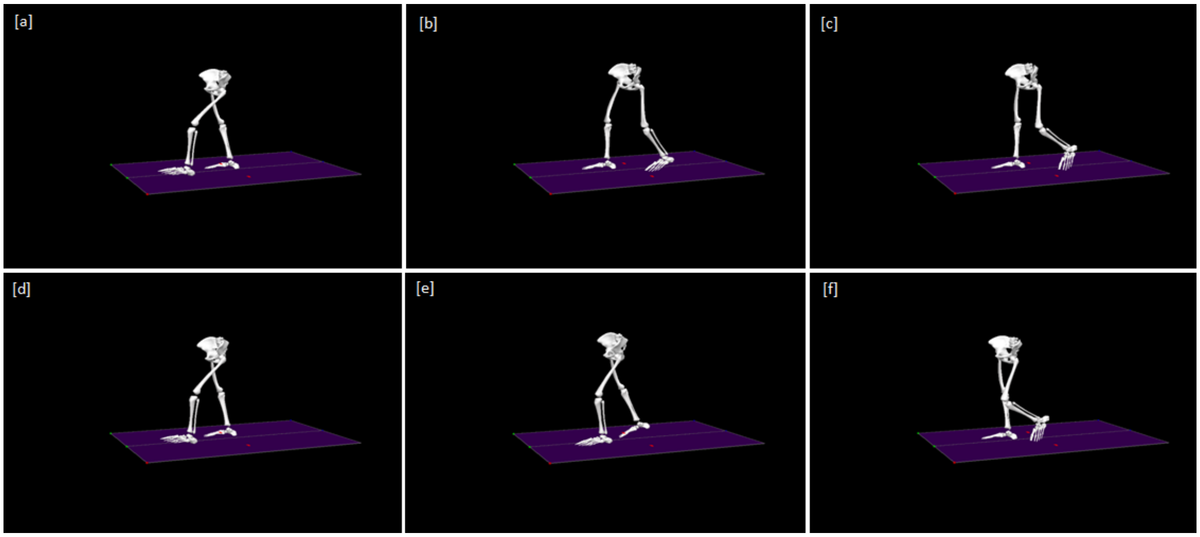


Figure 18: Visual 3D representation of lower body segments for OWS Participant 3 at a) PERTFS; b) REC1FS; c) PERTFO; d) REC2FS; e) REC1FO; and f) REC2FO.

4.2.6.2 Recovery 1 stride

In swing, leading up to REC1FS, knee extension velocity was greater than TL and knee flexion angle was less. Knee flexion was slightly greater at REC1FS but, through weight acceptance, knee extension velocity was much greater than TL, resulting in knee hyperextension in early stance (Figure 19). This contrasted with slight knee flexion following TL FS. Like participants 1 and 2, quick knee extension during weight acceptance was a preventative strategy to eliminate the threat of a knee collapse following the stumble.

At REC1FS, ankle plantar flexion velocity was less and was nearing zero, and ankle plantar flexion angle was slightly less than TL. Hip abduction, hip external rotation, pelvic obliquity, and negative pelvic rotation were slightly increased. Hip flexion and pelvic tilt were moderately increased compared to TL. Hip extension in stance was less than TL, while peak pelvic tilt angle was greater and occurred near REC1FS, opposed to during TL strides when the peak occurred near FO. Knee flexion velocity was less in late stance.

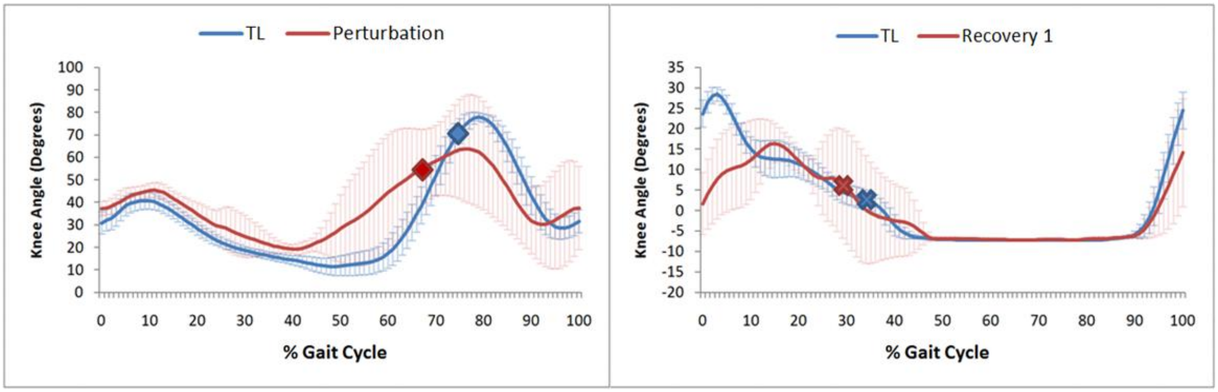


Figure 19: Knee flexion (+) and extension (-) angles for OWS Participant 3: from FS to FS for the left leg for TL and perturbation strides (left), and from FO to FO for the right leg for TL and recovery 1 strides (right).

At REC1FO, most joint angle values were similar to TL. Angles were slightly greater for hip external rotation and pelvic rotation, and knee flexion continued to be moderately less compared to TL FO values. Ankle plantar flexion velocity at REC1FO contrasted with dorsiflexion velocity at TL FO.

4.2.6.3 Recovery 2 step

At REC2FS, ankle dorsiflexion and knee flexion were greater compared to TL. Pelvic rotation was positive, which differed from TL FS, where pelvic rotation was negative. Knee extension velocity at REC2FS contrasted with flexion velocity at TL FS, and all other joint angular velocity values were similar to TL.

At REC2FO, ankle plantar flexion was less than TL and ankle adduction, hip flexion, and negative pelvic rotation were all slightly greater. Joint angular velocities were similar to TL at REC2FO.

4.2.7 OWS engagement and activity

During testing, no knee collapse stumbles occurred during gait on any treadmill surface condition. As a result, the OWS did not engage during TL gait, perturbed gait, or while walking on any of the other simulated surface conditions. The OWS SCKAFO engaged for one participant during CAREN accommodation trials, due to the participant increasing walking speed as they became more accustomed to walking in the SCKAFO. The angular velocity threshold was adjusted to accommodate the increased walking speed in these cases. Following the adjustment, the person let their braced limb collapse to ensure that the device would engage appropriately. All adjustments were made prior to any testing trials.

Mechanical issues with the OWS arose on two occasions due to leaks from the joint's fluid chamber. These issues occurred in the first two testing sessions. In the first instance, the testing session had to be rescheduled, and in the second, the leak occurred after the trials were complete. The leaks originated from ends of the chamber and once fluid was replaced and the brackets were tightened by an engineer at TOHRC, the issue did not arise again.

4.2.8 Questionnaire results

Questions concerning the comfort and aesthetic of the OWS SCKAFO revealed that one participant felt that the orthosis was comfortable, while two participants were only comfortable half the time (Appendix T). Participants found OWS appearance to be good to acceptable, though two participants commented that the joint shaft was large and stuck out; this could be an issue when wearing long pants. One person mentioned that if the pump leaked during daily use, as it did for his test, there would be a big mess.

In terms of walking and stability, participant 1 felt that the OWS orthosis was heavy, walking ability was worse, walking took more effort compared to his current orthosis, and the OWS orthosis did not improve support compared to walking without an orthosis. The other two participants found walking ability was improved, walking took less effort, stability was easy, and they felt very secure. Participant 3 found balance easy and commented during trials that he would not have been able to walk up the incline without the support of the OWS orthosis. Despite this category receiving some of the higher ratings reported in the questionnaire, anecdotal reports revealed that two participants felt the OWS SCKAFO was limiting when stepping forward (i.e. terminal stance).

Most negative comments toward the OWS SCKAFO were related to ease of use. All participants had trouble donning the OWS SCKAFO and required help from the research assistant. While free knee motion was rarely restricted by the OWS during walking trials, knee flexion restriction was an issue during sitting, since the current OWS design restricted the SCKAFO to 90° of knee flexion. While responses related to sitting were neutral-easy, all participants expressed that putting on a shoe was difficult or very difficult. Two participants required help from the research assistant to put on the shoe, while one used a shoe horn.

Overall, two people had a neutral attitude about wearing the orthosis, while one was accepting. Participant one felt that the OWS worked on and off, and didn't feel he could count on it. He also commented that fall situations he faces during daily activities typically occur very slowly. For example, as opposed to during walking, the leg may give out while shifting weight or turning around to do something, and since he tries to keep himself from falling, sometimes recruiting help from objects around him, the knee gives out quite slowly. For that reason, he commented that a velocity trigger "may not be the best way to go".

Chapter 5. Discussion

This research resulted in the development of a stumble perturbation in the CAREN-Extended virtual reality environment. While this method produced increased knee flexion upon weight acceptance for able-bodied participants, different landing strategies were employed by people with knee extensor weakness. This discussion will examine the outcomes for stumble perturbation evaluations with able-bodied participants and OWS evaluation with pervious KAFO users.

5.1 Stumble Development

The stumble method evaluation results indicated that a split-speed perturbation can evoke a knee collapse stumble in able-bodied participants. While two different recovery strategies were presented, knee flexion at REC1FS and the knee flexion peak following FS, were noticeably greater than TL gait for all participants and all trials.

Hopping and stopping recovery strategies were observed. These results were similar to Eng et al. (1994) who reported that, following a trip induced by a mechanical object in late swing, two different recovery strategies were used. Four subjects used several quick steps following the perturbation and one participant increased the length of the next step.

The hopping and stopping strategies were also consistent with stumble recovery strategies previously described by Cordero et al., (2003), who used a rope to restrict swing phase. Participants chose either a shorter step length and time, or a longer step length and time when these perturbations were applied in early swing. Cordero et al. concluded that participants either chose to complete the perturbed step and maintain speed or abort the perturbed step and rely more heavily on the contralateral limb for recovery, at the cost of slower speed. Though the perturbation types were different, the outcomes were similar; the hopping strategy showed shorter step lengths and maintained (or increased) speed while the stopping strategy involved halting gait, and resulted in longer step times and a loss of speed.

In this thesis knee flexion increased during weight acceptance for all participants, despite the two different stumble recovery strategies. Several studies also reported increases in knee flexion during swing and at FS after trip perturbations (DelMarcelle et al, 1997; Grabiner et al., 1993). Eng et al. (1994) also described flattened heel contact with a flexed knee upon landing.

Owings et al. (2001) reported that recovery mechanisms from treadmill acceleration were similar to those of mechanically induced tripping. Results from our study concurred with Owings and colleagues, implying that a split-speed perturbation produced similar stumble recovery outcomes as previous studies, without the use of mechanical objects.

Participant five used a ‘stopping’ strategy rather than the ‘hopping’ strategy, possibly due to limited accommodation time within the CAREN environment. Individual CAREN users may adapt to the surrounding virtual environment at different rates. Anecdotal comments by participant five during testing revealed a general sense of unease toward treadmills. This participant used the handrails for security during walking and recovery.

Self-paced treadmill mode was used in this protocol to reproduce the most natural stumble recovery tendencies. The ‘stopping’ strategy observed supports this decision since, if a constant treadmill speed was used, the person would have a rolled off the back of the treadmill. A different recovery strategy would need to be adopted in order to avoid a fall, which may not accurately reflect the individual's natural tendencies. Self-paced mode allowed the person to quickly slow down and recover before resuming regular walking.

Ankle, knee, hip, and pelvis angles were similar between TL FS events preceding PERTFS. This indicates that participants did not predict the perturbation prior to its initiation, and therefore did not alter their gait pattern before the perturbation occurred. Therefore, the split-speed perturbation may mimic real world stumble situations more effectively than methods in the literature where the stumble may have been anticipated.

Differences in angular velocities on TL were consistent with findings from Winter et al. (1987), since participants 1-4 walked faster and had higher absolute angular velocity values at the ankle, knee, and hip. Participant 5 walked slower and had lower absolute angular velocities. Differences in angular velocity between the hopping and stopping strategies also agree with Winter's findings, with the hopping strategy producing higher absolute velocities during perturbation and recovery 1 strides (speed increased through recovery) than the stopping strategy (speed decreased through recovery). Exceptions at certain FS angles were mostly explained by the crouching motion when stopping, where some angular velocity values increased to propel the body upward to a standing position.

Recent research has shown that, with repeated trip or slip training, recovery strategies can be improved and retained for several months, suggesting that training using induced slip or trip methods could be beneficial in clinical rehabilitation programs (Bieryla, 2007; Pai & Bhatt, 2007; Parijat, 2011). The ability to adjust perturbation settings to alter intensity and timing makes this application accessible to populations and individuals who may react differently to the perturbation. This method could be used in CAREN labs or other real-time controlled dual-tread treadmill systems, although the CAREN-Extended system allows for stumble perturbations during a wide variety of ground orientations and scenarios. Since the split speed stumble method used for this project was created, the perturbation tool has become part of regular clinical rehabilitation practices within the RVR Lab.

5.2 OWS Evaluation

5.2.1 Stumble recovery

While OWS users stumbled following split speed perturbation, the ensuing stumble did not increase knee flexion during weight acceptance and did not lead to the desired knee collapse. Though kinematic results showed more knee flexion at REC1FS than at TL FS, these participants with knee extensor weakness extended their leg before and after REC1FS to respond to the threat of knee collapse. Instead of increased knee flexion during weight acceptance, as hypothesized, OWS participants actually extended the knee faster and, in almost all cases, more extended knee angles were reported following stumble situations. Only participant 3 did not have greater knee extension at weight acceptance, displaying similar knee flexion values through weight acceptance in TL and recovery 1 strides. Participant 3 already hyperextended his knee during stance in TL walking, which explained why the knee was not more extended compared to TL in recovery. Negligible increases to knee flexion angles were seen in only two trials, by 0.05° in trial two and in the avoided trial.

Unexpectedly, the stumble perturbation that caused increased knee flexion for able-bodied participants did not produce the same effect for people with knee extensor weakness. Instead, knee extension occurred following REC1FS, even through the participants had knee extension deficits. A kinetic data analysis would provide knowledge of the biomechanical mechanism with which the lower extremity kinetic chain contributed to this knee extension; however, knee flexion kinematics provided a description for how OWS participants were able to quickly straighten the leg following REC1FS.

For unperturbed TL walking, knee flexion angular velocity at FS was -161.64 ± 40.30 for able-bodied participants, but knee angular velocity for all OWS participants was much lower (-14.02 ± 25.3) and remained conservative throughout stance. Participant 3 even displayed a knee extension angular velocity at TL FS (i.e., knee was extending at foot strike). Considering that people with quadriceps weakness tend to keep the knee straight during stance phase, as shown in the results, knee flexion velocity was expected to be closer to 0 to prepare for knee extension (Siegel et al., 1993).

Following the perturbation, knee angular velocity differences between able-bodied and OWS participants at REC1FS were even greater. Able-bodied participants had an average knee flexion velocity of -62.88 ± 104.89, indicating further flexion through weight acceptance. However, OWS participants had a knee extension velocity of 128.30 ± 53.86, indicating quick knee extension through weight acceptance. In fact, the knee was already extending before REC1FS (Appendix O). From figures 14 and 16, participants 1 and 2 landed with their leg in front of the pelvis, such that the total body centre of gravity would promote knee extension. Kinetic analysis would be required to confirm this assumption. Participant 3 was close to full extension at REC1FS.

The knee extension velocity at REC1FS and early stance played a role in the OWS participant’s ability to avoid knee flexion that could threaten a knee collapse, despite quadriceps strength deficits. These experienced KAFO ambulators used free moving knee joints on their regular orthosis and likely had experience landing from stumbles with a straight leg. Participants continued to use this gait strategy during the stumble.

Another possible reason that the stumble method was not effective in invoking a knee collapse for the three OWS participants is that their muscle strength of -4 to 4 could have helped then initiate extension before weight-bearing (i.e., during swing following perturbation). Additionally, hip flexor strength ranged from 4 to 5 out of 5, enhancing their ability to position the leg and extend the knee in the period around REC1FS. Adaptations to normal gait have been found to be more effective in those who are stronger in the uninvolved musculature, and this finding would also apply to stumble situations (Siegel et al., 1993). The split speed perturbation may have been more effective for invoking knee collapse for participants with lower knee and hip strength, since they may not have the ability to bring a straightened leg forward as quickly for recovery.

The use of handrails may also explain how OWS participants were able to recover from the stumble perturbation. While handrails were only used for recovery in the stopping strategy for able-bodied participant 5, all OWS participants used handrails for support periodically during gait and during stumble recovery. Upon review of digital video, the handrails appeared to help participants briefly prop themselves up following the perturbation. This allowed extra time for the recovery leg, which was in swing, to prepare for REC1FS. Using handrails to help position the braced leg mainly applied to Participants 1 and 2, who were faster walkers. In these cases, the stumble method may have been more effective if the handrails were not available. Upper body kinetic data would have provided a more thorough understanding of handrail use for stumble recovery.

With the recovery strategies used by OWS participants, the OWS joint did not reach the threshold angular velocity to engage. The OWS would not be required for similar stumbles during walking gait for the tested population. However, the OWS could be effective for preventing knee collapse in other stumble situations.

One participant commented that most of his stumble situations during daily activities occurred quite slowly, usually during situations involving irregular weight transfer, such as turning around from a standing position. When the participant detects instability during these situations, the initial reaction is to prevent a fall in any way possible, including grabbing surrounding objects for support. In these cases, the fall occurs quite slowly. The speed of the falling body and resulting knee flexion angular velocity would likely be slowed too much to surpass the angular velocity threshold, which is set according to maximum knee flexion velocity during gait. Therefore, the OWS would not engage to prevent a knee collapse and would not be of additional benefit as a fall prevention aid.

To enable to OWS to engage for these daily activity knee collapse events, additional training would be required. If users could learn to trust the joint, allowing themselves to fall quickly in these situations and let the OWS catch them, the device would be useful for preventing knee collapse. For people who have already developed compensatory strategies to prevent themselves from losing stability during daily activities, a change in gait paradigm would be required by the user to modify existing instinctive reactions and let themselves collapse onto the OWS joint. This would likely take more time for experienced KAFO users who have difficulty adapting to a SCKAFO due to their ingrained KAFO gait strategies (Davis et al., 2010; Irby et al., 2007).

Out of all eight participants, OWS participant 3 was the only one who displayed joint angle differences at PERTFS compared to TL FS. OWS participant 3 had the slowest walking speed and the perturbation length (time that treadmill speed was altered) had to be increased. The increased perturbation time meant that the change in treadmill belt speeds occurred over a longer time, before foot contact at PERTFS. The participant may have anticipated the upcoming perturbation by visually or audibly detecting the treadmill change, which could have resulted in gait adjustments prior to prepare for PERTFS. If anticipation occurred, this may also explain why OWS participant 3 was able to avoid one of the perturbations by aborting the perturbation stride during these trials.

Preferably, the stumble perturbation method would have been developed with KAFO user population. Participants with similar functional deficits would likely have similar recovery responses that differ from able-bodied participants. However, recruitment issues for KAFO users for research studies, fatigue issues with prolonged development trials, and injury risk during development provide a rationale for developing the split-speed perturbation method with an able-bodied population.

5.2.2 OWS engagement and activity

Because the stumble method used in OWS testing did not cause knee collapse, parameters related to the timing and flexion angles for OWS engagement could not be evaluated. However, OWS testing did show that the device did not engage to resist knee flexion during TL walking in testing trials, as hypothesized. Free knee motion was also achieved for the other surface conditions. Since the OWS's engagement threshold was based on individual walking speed, participants were able to continue their TL gait patterns without disruption by the joint. This is an important factor for successful SCKAFO designs, since free knee motion produces near normal knee flexion during swing phase and reduces the chance that people with quadriceps weakness will develop undesirable compensatory strategies (Herbert & Liggins, 2005; Irby et al., 2007; Irby et al. 1999). Anecdotal feedback supported the importance of free knee motion during gait, since OWS participants admitted to becoming frustrated in the past, and removing elements of their current orthosis that restricted knee flexion. This feedback implies that free knee motion is important to device satisfaction for some orthosis users at a higher functional level. Higher function KAFO users require some support but may be physically able to walk without an orthosis in some conditions and are more likely to abandon their orthosis than someone who requires their orthosis to walk (Hong, San Luis & Chung, 1990).

Mechanical issues arose twice during testing, when the joint leaked hydraulic fluid. Though the prototype OWS was fixed very quickly by the engineers by replacing fluid and tightening seals, these issues would need to be address in commercial versions of the device.

5.2.3 OWS SCKAFO questionnaire

Mixed feelings were observed in regards to the comfort and aesthetic appeal of the orthosis. However, this result could be improved with more extensive fitting processes that were not pursued for the purposes of this project. In terms of cosmesis, though OWS appearance was rated as acceptable or good, comments revealed that participants did not appreciate the size and shape of the joint shaft, and felt that it may be an issue while wearing long pants. The participants were used to small single-axis joints (smallest knee joint on the market, providing only free motion) for their current KAFO.

The best ratings related to walking and stability. Participants were able to adapt quickly to walking in the SCKAFO, since all individuals were used to walking with free knee motion in their own orthoses. This implies that current SCKAFO users may not have trouble adjusting to the OWS SCKAFO. However, this may not representative of how experienced KAFO users with locked knee joint would adapt (Irby et al., 2007).

Participants also expressed that their walking ability was especially improved when walking uphill. This could be related to the strength of the carbon fiber used for the OWS orthosis, as compared with their current thermoplastic KAFO. The carbon fiber laminate shell is stiffer, thinner, and lighter, but the lower weight gains are countered by stronger and heavier steel uprights and a heavier OWS joint. The gains for walking stability on inclines may be worthwhile for some orthosis users.

Most negative comments were related to ease of use. The current OWS design restricts the SCKAFO to 90° of flexion, which was an issue during sitting. The ability to don/doff the orthosis and put on/take off the shoes independently was affected by this angle, and in some cases participants required assistance from the research assistant during these tasks. These tasks normally occur multiple times per day and the ability to perform them independently is important, since ease of use has been shown to affect rates of user satisfaction with assistive devices (Bernhardt et al., 2006; Phillips & Zhao, 1993). The flexion restrictions on the current OWS design should be addressed before being released to the consumer market.

Chapter 6. Conclusion

6.1 Stumble Development

Two stumble recovery responses were observed in able-bodied participants following the split speed perturbation: hopping and stopping. Increased knee flexion at FS and in weight acceptance was achieved following the split-speed perturbation, for both recovery responses, in all trials, for able-bodied participants. Knee flexion increased by 16.74° at REC1FS and 16.36° in weight acceptance for participants using the hopping strategy, while knee flexion increased 32.29° at REC1FS and 76.58° in weight acceptance for the participant using the stopping strategy.

6.2 OWS Evaluation

Contrary to our hypothesis, the split speed perturbation increased knee flexion during weight acceptance for able-bodied participants but did not induce increased knee flexion during weight acceptance for orthosis users. Instead, a straight-legged response was employed. Increased knee flexion was achieved at REC1FS compared to TL FS, with increases of 8.38° for Participant 1, 9.88° for Participant 2, and 3.43° for Participant 3. However, the knee extended following REC1FS, resulting in similar or greater knee extension during recovery 1 weight acceptance. Participants 1 and 2 had greater knee extension angles, by 9.16° and 25.71°, respectively. Participant 3 had knee hyperextension during recovery 1 weight acceptance, which was similar to TL weight acceptance. Since the straight legged response did not invoke increased knee flexion during recovery 1 weight acceptance, as desired, the split speed perturbation was not appropriate to measure OWS engagement response.

The hypothesis that OWS would allow free knee motion during gait was accepted following review of knee joint kinematic data, digital video, and manual recordings. The OWS did not engage during gait on TL surfaces or following the split speed perturbation.

Feedback related to walking ability in the OWS was positive. Concerns that need to be addressed in future OWS prototypes include mechanical issues due to fluid leaks, and limitations during daily tasks due to knee flexion restriction to 90° in the current prototype.

6.3 Future Research and Development

6.3.1 Stumble Development

Due to the small sample size in this exploratory study, inferential statistical analysis was not pursued. Future research incorporating this stumble method could be useful to assess stumble recovery strategies of a larger group of able-bodied individuals, people with disabilities, or the elderly.

6.3.2 OWS Testing

Observations during digital video analysis suggested that handrail use allowed extra time for OWS participants to prepare for REC1FS. The split speed perturbation may have been more effective for inducing knee collapse in OWS participants if the handrails were removed from the platform. However, handrail removal may have also introduced a more protected gait strategy. Future testing is needed to confirm the effects of handrail removal.

In this study, the OWS was not required during recovery, but may assist recovery in other stumble situations causing knee collapse. Random knee collapses that occur during daily life are difficult to simulate in a controlled research setting. In the future, the OWS could be tested outside the laboratory, with reports from orthosis users to determine OWS response to real world stumble situations. Additionally, Participant 1 suggested that most of his knee collapse situations occur during irregular weight transfers, and occur very slowly. Further inquiry toward typical fall situations within the OWS target population should be carried out, and results should be considered for future OWS testing and training.

Mechanical issues and limitations related to tasks such as putting on shoes and donning and doffing the orthosis impact orthosis user satisfaction. Further research should confirm that these issues are resolved in future OWS prototypes, before the device is released to a consumer market.

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Appendix A: Stumble Development Recruitment Notice

Functional and biomechanical assessment of an angular-velocity approach for stance control orthoses in a virtual reality environment

Dear Sir/Madame:

The Ottawa Hospital Rehabilitation Centre (TOHRC) is testing a new device called the Ottawalk-Speed joint (OWS). This device was designed by staff at TOHRC and has the potential to improve upon current orthotic designs, which are used by people with lower limb weakness. This research aims to assess the performance of the OWS. In order to properly test this device, a pilot project must be completed. Results from pilot testing will allow the research team to design a stumble-inducing method that will be used during OWS testing. Able-bodied participants are being recruited for the pilot portion of the research.

If you choose to participate in this study, you will be asked to complete one testing session in the Rehab Virtual Reality (RVR) Lab, which houses the Computer Assisted Rehabilitation Environment (CAREN). The CAREN is a virtual environment with a treadmill, a moving platform and a projection screen. After a tour of the Lab, reflective markers will be attached to your body and you will be secured to a safety harness. The harness will be attached to an overhead structure and will prevent you from falling while immersed in the virtual environment. To ensure accurate data collection, you will be required to bring tight-fitting, non-reflective clothing to wear during the session. You will walk through “The Park” virtual scene once to become accustomed the environment. Following the accommodation trial, you will walk through “The Park” while the operator perturbs the scene in an attempt to make you stumble. Movements of the visual field and platform, changes to the treadmill speed and multi-tasking activities may be used to build the stumble method. If at any time the research team deems a situation unsafe, Emergency Stop procedures will be initiated and the platform and treadmill will stop immediately. In this event, you will be taken out of the CAREN and may resume testing if and when you are comfortable doing so. The session will take about an hour to complete.

To participate in this study, you must be over 18 years of age and must not possess any significant physical deformity or disability. You must also be in good enough physical condition to withstand multiple stumble situations.

Participation is on a volunteer basis; you are not obligated to participate. If you decide to participate, you will be free to withdraw from the study at any time without suffering any negative consequences and without it affecting any of your present or future relationships with TOHRC. All testing will take place in the RVR Lab at TOHRC.

Please reply to this email if you are interested in participating, if you have any questions, or if you would like more information to assist you in making your decision.

Thank you for your time,

The Ottawa Hospital Rehabilitation Centre

Appendix B: Patient CAREN Orientation and Procedure Form

The Ottawa Hospital | L'Hôpital d'Ottawa

Principle Investigator:

Welcome to the RVRL. To make your sessions successful, please adhere to the following instructions:

Bring tight fitting, non-reflective clothing to wear during your test session. Please do not wear jewelry.
Bring gym shoes that are non-reflective and non-marking to wear during your test session.
Secure long hair close to your head.
Be aware of and follow the safety rules of the RVR Laboratory at all times. More information regarding the responsibilities of the researcher and your responsibilities as a participant is provided in the following pages of this handout. Please review this information before attending your test session in the RVR Lab.
Be on time for your appointment.
Please notify your researcher in a timely manner if unable to attend your testing session.

Session Information:

The RVRL is located in Room 1505, first floor of the Rehabilitation Centre.

Date:

Time:

Additional Information

For your information, relevant sections of the CAREN Operating Manual have been included in this document. Please review this information carefully, and discuss any questions or concerns with the research assistant before attending your session in the RVR Lab.

6.0 SAFETY REGULATIONS

6.1 RESEARCH SESSIONS

6.1.1 No one will be allowed to use the CAREN system without a certified operator present and a second person (researcher) present.
6.1.2 No one will be allowed onto the platform unless initially accompanied by the operator or attending researcher.
6.1.3 No one (patients, research participants, staff, visitors, researchers) is allowed on the CAREN platform without a safety harness, while the system is in use.
6.1.4 The attending researcher and operator are responsible for ensuring that the harness is properly secured.
6.1.5 The attending researcher and operator are responsible for ensuring that the tether rope is the correct length.
6.1.6 All persons on the platform should be facing the screen
6.1.7 No one will get on and off the platform without using the bridge.
6.1.8 Authorized spectators will be allowed to attend clinical/research sessions at the discretion of the attending clinician/researcher and will be asked to stay in a specified area.

7.0 ROLES & RESPONSIBILITIES

RESPONSIBILITIES OF THE PATIENTS USING THE RVR LABORATORY
The patient shall:

- Wear snug fitting, non-reflective clothing during session and should not have loose hair or jewelry.
- Wear either gym shoes or other pliable non-marking, non-reflective footwear, to assure firm grip and avoid damaging the treadmill.
- Secure long hair should close to the head.
- Be accompanied by their attending (researcher)
- Be aware of and follow the safety rules of the RVR Laboratory at all times.

RESPONSIBILITIES OF THE RESEARCHER
The researcher is responsible for:

- The safety and well being of the research participant while in the laboratory and ensuring that that the use of the CAREN system is appropriate for their research participant.
- Reviewing the protocol with operator to plan the specific session expectations, prior to the initial research CAREN session.
- Either stand on the platform with the subject during a session, where he/she can see the patient and reach the emergency button at the same time or they must stand outside of the system's operating area during the session, where he/she can see the patient and reach the emergency button at the same time. If the researcher is on the platform they must also wear a safety harness.
- Reviewing CAREN safety protocols with research participant prior to or at the initial visit, with the operator.
- Following all applicable RVR Laboratory procedures
- Ensuring that safety and cleanliness of the RVR Laboratory are maintained and reporting any concerns regarding the RVR Laboratory to the supervisor.

Research projects involving the use of the RVR Laboratory or the Operator must follow standard IRRD and TOH research policies and procedures.

8.0 DRESS REGULATION

8.1 PATIENT/PARTICIPANT DRESS CODE

8.1.1 While on the platform, patient/participants must wear snug fitting clothes to avoid getting caught in the treadmill, and with not reflective materials as to help with accurate VICON marker placement, and motion measurements. Shorts or pants are appropriate for patients and research participant.

8.1.2 They must tie up hair, and wear no dangling clothes or accessories

8.1.3 They must wear clean athletic shoes, with a good non-marking tread, and with no reflective markings

Appendix C: Stumble Development Information and Consent Form

The Ottawa Hospital | L'Hôpital d'Ottawa

Information Sheet and Consent Form
Functional and biomechanical assessment of an angular-velocity approach for stance control orthoses in a virtual reality environment

Principal Investigator:
Funding Source: OTTN, NSERC

Introduction

You are being asked to participate in this research project to help the research team create a stumble-inducing method, which will be used for the primary research goals. Able-bodied people will be recruited for this pilot project.

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

Background, Purpose and Design of the Study

Lower limb orthoses are typically used by populations with quadriceps weakness to improve user safety and function. Engineers at The Ottawa Hospital Rehabilitation Centre (TOHRC) have developed a new approach for stance control orthoses called the Ottawalk-Speed joint (OWS). The device offers support to the user to prevent a knee collapse. People with limited or short-term quadriceps weakness may benefit from this design. This research will test the OWS on orthosis users and assess device performance.

In order to assess the performance of the device, researchers must determine a method that will cause OWS testing participants to stumble. The stumble method will be developed during a pilot portion of the research using the Computer Assisted Rehabilitation Environment (CAREN), located in the Rehab Virtual Reality (RVR) Lab at TOHRC. The CAREN contains a treadmill, a moving platform and a projection screen. Five healthy people will be recruited to complete a testing session in the RVR Lab. This session will be recorded with digital video in order to time match data. Results from these sessions will be used to develop a stumble-inducing method, which will be used during OWS testing.

<u>Study Procedures</u>

You will complete a testing session in the RVR Lab at TOHRC. The complete protocol is listed below:

- A certified operator will give you an orientation to the CAREN system and describe safety procedures, including Emergency stop procedures.
- Four reflective markers will be placed on your torso on tight-fitting, non-reflective clothing. A marker will also be placed on the back of each of your hands.
- You will be fitted to a safety harness and escorted to the CAREN platform. You will be secured to a safety tether attached to the fall arrest overhead structure.
- You will complete one walking trial in the 'Forest Road' scenario to adjust to the virtual environment.
- You will walk through the 'Forest Road' while the operator uses strategies to attempt to make you stumble.
- Strategies that the operator may use to create a stumble include:
    - Visual field changes: The visual field may be altered in an attempt to confuse your view of the virtual scene.
    - Platform movements: The platform can be moved forward, backward, upward, downward, or to either side at various speeds. It can also be rotated in any direction. These movements can be applied individually or together in an attempt to throw off your balance.
    - Treadmill speed changes: Treadmill speed can be rapidly altered within the range. Rapidly changing treadmill speed may throw off your balance and create a stumble situation.
    - Dual-task activities: The Forest Road application includes dual-task mode that requires you to use your arms to 'hit' stationary or moving objects. When you must focus on two tasks, it may be more difficult to keep your balance.
- If Emergency Stop procedures are carried out at any time during the session, you will be taken off the CAREN platform. The session will be repeated if and when you are comfortable.
- When CAREN testing is finished, you will be escorted off of the CAREN platform. Markers and the harness will be removed and the session will be complete.

Study Duration

The entire study is expected to take 12 months to complete. You will participate in one testing session during the study. Total session time will be approximately one hour in length, or will last until the research team finds a perturbation method causing you to stumble in 5 consecutive trials. Once this session is complete your role in this study will be finished.

Possible Side Effects and/or Risks

This research is minimally invasive and it is unlikely that any serious injury will occur. Irritation may be caused by the CAREN equipment in the event that the fall arrest harness is abruptly employed to support you to prevent a fall.

Though the CAREN will disrupt your balance due to platform and treadmill movements during stumble trials, these trials will cause no more risk to injury than a stumble that could occur in an everyday walking situation. In fact, stumble trials will be more safely executed on the CAREN than in a normal floor due to the support of the safety bars and harness. Additionally, Emergency Stop procedures will be executed immediately in any situation deemed unsafe by the operator or research assistant.

Benefits of the Study

You may not receive any direct benefit from participating in this study. Your participation will allow the research team to assess the performance of a new stance control design, which may be beneficial to orthosis users in the future.

Withdrawal from the Study

You have the right to withdraw from the study at any time without any impact to your current and future association with the Ottawa Hospital or The University of Ottawa. If you decide to withdraw, you should discuss this with the primary investigator before you stop the study. You will also have the option to withdraw any of your data collected during the course of the study. You may be withdrawn from the study if you do not follow study staff's instructions.

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

Compensation

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

Confidentiality

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

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

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

Voluntary Participation

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

Questions about the Study

If you have any questions about this study, or if you feel that you have experienced a research-related injury, please contact.

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

Consent Form
Functional and biomechanical assessment of an angular-velocity approach for stance control orthoses in a virtual reality environment

Consent to Participate in Research

I understand that I am being asked to participate in a research study to build a stumble–inducing method that will be used for OWS assessment. This study has been explained to me by the research assistant.

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

I voluntarily agree to participate in this study.

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

Signatures

____
Participant's Name (Please Print)

________________________________ __________________
Participant's Signature Date

Investigator Statement (or Person Explaining the Consent)

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

____
Name of Investigator/Delegate (Please Print)

______________________________________ __________________
Signature of Investigator/Delegate Date

Appendix D: OWS Recruitment Notice

The Ottawa Hospital | L'Hôpital
Hospital | d'Ottawa

Functional and biomechanical assessment of an angular-velocity approach for stance control orthoses in a virtual reality environment

Dear Sir/Madame:

The Ottawa Hospital Rehabilitation Centre (TOHRC) is testing a new device called the Ottawalk-Speed joint (OWS). This device was designed by staff at TOHRC and has the potential to improve upon current orthotic designs while offering support to people with lower limb weakness. Persons who are currently prescribed to use an orthosis are being asked to participate in a research study that will assess the performance of the OWS during normal walking and stumble situations.

If you choose to participate in this study, you will be asked to come in to TOHRC to complete OWS fitting and training and one testing session. In the OWS fitting and training, you will be fit to the OWS by an orthotist. The fitting time will be similar to that of your current orthosis. After successful fitting, you will practice walking and knee collapse situations within safe areas of the TOHRC gym until you are comfortable. Training sessions will be no longer than 1.5 hours in length. At this time, a consensus that you are ready to move on to the testing session must be reached between you, the participant, the orthotist, and the research team. Multiple training sessions may be required if a consensus is not reached. Your involvement in the study may be reassessed by the orthotist if a consensus is not reached after multiple sessions.

The OWS testing session will be completed in the Rehabilitation Virtual Reality (RVR) Lab, which houses the Computer Assisted Rehabilitation Environment (CAREN) system. The CAREN is a virtual environment with a treadmill, a moving platform and a projection screen. After a tour of the Lab, reflective markers will be attached to your body and the OWS orthosis and you will be secured to a fall-arrest harness. This harness will be attached to an overhead structure and will prevent you from falling while immersed in the virtual reality environment. In order to achieve accurate data collection, you will be required to bring tight-fitting, non-reflective clothing to wear during the test session. The operator will disturb the virtual scene while you walk in the "The Park" in an attempt to cause a stumble. Five stumble trials will be collected in total in order to assess how the device behaves during a stumble. The safety harness will prevent you from falling during stumble trials. The Emergency Stop sequence will be executed if the research team deems the situation unsafe at any time during testing, which will cause the platform and treadmill to stop immediately. In the event that an Emergency Stop is initiated you will be taken out of the CAREN, and testing will resume if/when you are comfortable. Breaks will be given as needed. When testing is completed, you will be asked to complete a questionnaire. This session will be approximately 2 hours in length.

To participate in this study, you must be a current orthosis user, be 18 years of age or older, weigh less than 90 kg/200 lbs, and be able to walk continuously for 80 metres/260 feet. You must not possess any ailment (other than lower limb weakness) that will prevent you from safely completing testing. Participant inclusion will be determined by the above criteria as well as by the decision of the clinician (orthotist/physiatrist). In order to participate in this research, the clinician must agree that you have the ability to stabilize yourself while walking with the OWS.

Participation is on a volunteer basis; you are not required to participate. If you decide to take part in the research, you will be free to withdraw from the study at any time without any effect to your present or future status with TOHRC. All sessions will take place at TOHRC. Parking expenses will be reimbursed by the research team.

If you would like to participate, have any questions, or if you would like more information to assist you in making your decision, please respond using the contact information listed below:

In the event that the research assistant is not contacted after two weeks of sending this letter, a follow up telephone call will be made to ensure you have received this letter.

Thank you for your time,

The Ottawa Hospital Rehabilitation Centre

Appendix E: OWS Information and Consent Form

The Ottawa Hospital | L'Hôpital d'Ottawa

Information Sheet and Consent Form
Functional and biomechanical assessment of an angular-velocity approach for stance control orthoses in a virtual reality environment

Principal Investigator:
Funding Source: OTTN, NSERC

Introduction

You are being asked to participate in this research project because you are a current knee ankle foot orthosis or stance control knee ankle foot orthosis user. This research project will assess a new stance control design and must be tested on current orthosis users.

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

Background, Purpose and Design of the Study

Engineers at The Ottawa Hospital Rehabilitation Centre (TOHRC) have developed a new approach for stance control orthoses called the Ottawalk-Speed joint (OWS). The device offers support to the user to prevent a knee collapse. People with limited or short-term quadriceps weakness may benefit from this design. This research will test the OWS on orthosis users and assess device performance.

The goals of this research are to: 1) Determine if the device prevents a knee collapse, 2) Assess how the device reacts in stumble and normal walking situations, and 3) Assess user opinions of the device. Five current orthosis users will complete a testing session in the Rehab Virtual Reality (RVR) Lab at TOHRC. The RVR Lab houses the Computer Assisted Rehabilitation Environment (CAREN). The CAREN is a virtual environment containing a moving platform with a mounted treadmill, a projection screen, and various tools for measuring movement. Testing sessions in the RVR Lab will be recorded with digital video in order to time match data. Results from the RVR sessions will be used to assess OWS performance, and questionnaire results will be used to assess ratings of user satisfaction with the OWS.

Study Procedures and or Description of Treatment

You will complete OWS fitting and training and one testing session at TOHRC. The complete protocol is listed below:

OWS Fitting and Training

- The orthotist will fit you to the OWS SCKAFO. This will be similar to previous fittings with your current orthosis.
- You will be instructed on use of the orthosis and will be trained by the orthotist until you feel comfortable with the device and can successfully engage the OWS during a stumble. Training activities include:
    - Walking between parallel bars
    - Walking beside a spotter
    - Practicing knee collapse situations between parallel bars
    - Walk safely in the gymnasium
- Consensus between you and the orthotists must be reached before moving onto OWS testing. The time to complete this phase is unknown at this time since this is the first pilot study with the OWS and OWS is a unique device (i.e., no literature on use by people with disabilities).
- Once the session is completed you will return to your original orthosis, and will schedule a time for OWS testing with the research assistant.

OWS Testing

- You will be fit to the OWS used during the 'OWS Fitting and Training' session(s) by the orthotist.
- In the RVR Lab, the system operator will give you a brief orientation to the CAREN system, and describe safety procedures including emergency stop procedures.
- You will prepare for testing:
    - You will wear tight-fitting, non-reflective clothing and will be fitted to a safety harness. Reflective markers will be attached to your body/clothing, the harness, and the orthosis of your braced leg.
    - You will be escorted to the CAREN platform and secured to a tether attached to the overhead structure. Safety hand rails will be attached to the platform during testing.
- You will complete a brief system acquisition in order to become familiar with the virtual environment. The acquisition session will last approximately 10 minutes and will be conducted as follows:
    - You will walk through the 'Endless Road' scenario at your own pace until you feel comfortable, for up to 5 minutes. You will be introduced to the 'Endless Road' first because it is the least challenging scenario offered by the CAREN.

    - You will walk through ‘The Park’ scenario at your own pace until you feel comfortable, for up to two cycles of the Road. ‘The Park’ is more challenging than the ‘Endless Road’, and will present you with various slopes and curves along the path.

- Five stumble trials will be collected. While you walk through ‘The Park’, the CAREN operator will apply the stumble method on level-ground sections. Stumble trials will be approximately 1-2 minutes apart.
- After the RVR Lab session, you will be taken to the RVR Lab’s conference room fill out a questionnaire assessing your level of satisfaction with the OWS.

Study Duration

The entire study is expected to take 12 months to complete. Over the course of the study, you will be involved in OWS fitting and training and one testing session on separate days. Time for OWS fitting and training will be similar to that of your current orthosis. Sessions will not exceed 1.5 hours in length, though multiple sessions may be required. The OWS testing session will be no longer than 2 hours in length. Once these two sessions are complete, your role in the study will be finished.

Possible Side Effects and/or Risks

This research is minimally invasive and it is unlikely that any serious injury will occur. The OWS SCKAFO may cause minor irritation similar to that caused by your current orthosis. Minor irritation may also be caused by the CAREN equipment in the event that the fall arrest harness is abruptly employed to support you to prevent a fall.

Though the CAREN will disrupt your balance due to platform and treadmill movements during stumble trials, these trials will cause no more risk to injury than a stumble that could occur in an everyday walking situation. In fact, stumble trials will be more safely executed on the CAREN than in a normal floor due to the support of the safety bars and harness. Additionally, Emergency Stop procedures will be executed immediately in any situation deemed unsafe by the operator or research assistant.

Benefits of the Study

You may not receive any direct benefit from participating in this study. Your participation will allow the research team to assess the performance of a new stance control design, which may be beneficial to orthosis users in the future.

Withdrawal from the Study

You have the right to withdraw from the study at any time without any impact to your current and future care at the Ottawa Hospital. If you decide to withdraw, you should discuss this with the primary investigator before you stop the study. You will also have the option to withdraw any of your data collected during the course of the study.

The study orthotist may request you withdraw from the study if they determine that this study is not right for you. You may also be withdrawn from the study if you do not follow study staff's instructions.

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

Compensation

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

Study Costs

You will be reimbursed for parking costs acquired during testing sessions.

Confidentiality

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

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

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

Voluntary Participation

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

New Information About the Study

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

Questions about the Study

If you have any questions about this study, or if you feel that you have experienced a research-related injury, please contact.

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

Consent Form
Functional and biomechanical assessment of an angular-velocity approach for stance control orthoses in a virtual reality environment

Consent to Participate in Research

I understand that I am being asked to participate in a research study that will assess the performance of the OWS. This study has been explained to me by the research assistant.

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

I voluntarily agree to participate in this study.

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

Signatures

____
Participant's Name (Please Print)

________________________________ ____________________
Participant's Signature Date

Investigator Statement (or Person Explaining the Consent)

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

____
Name of Investigator/Delegate (Please Print)

______________________________________ ____________________
Signature of Investigator/Delegate Date

Appendix F: Participant Information Sheet

Participant Code	
Casting Date	
Age	
Gender	☐M ☐F
Weight	
Current Orthosis Type	
Years of Orthosis Use	
Reason for Orthosis Prescription	

Manual Muscle Test

	Flexors	Extensors
Hip		
Knee		
Ankle		

Appendix G: Casting Session

An orthotist took a negative cast of the braced leg for each potential OWS participant. A tensor sock was first applied to the bare leg to protect the skin. Outlines of anatomical landmarks (i.e. femoral epicondyles, malleoli, patella, etc.) were drawn directly onto the sock, to be used as a reference during fabrication. Gypsona plaster bandages were dipped in water before being wrapped around the affected leg. With the leg fully extended, the first application started under the knee and around the ankle, and a second application was wrapped around the knee the knee to the top of the thigh. Once dried, the cast form was cut from the leg, and the cut section was rejoined to create a finished negative cast.

The negative cast became the mould to create a positive cast of the participant's leg. The positive cast was subsequently used by orthotists and technicians to create custom orthoses for each of the three eligible participants. For the initial testing of the OWS, the SCKAFO's structure was fabricated using PVC covered in a carbon fibre braid and acrylic resin. Joint uprights were attached within the frame of the orthosis, which makes the orthosis more cosmetically appealing and strong than outside attachments. For consistencies sake, each of the OWS SCKAFOs included an ankle plantar stop; plantar stops are used on SCKAFOs to restricted excessive plantar flexion, and can be individualized to not be restrictive during gait.

Appendix H: OWS SCKAFO Questionnaire

OWS SCKAFO Questionnaire

Date (dd/mm/yy)	

Participant Code		Age		Gender	☐M ☐F
Orthotist		Diagnosis			
Current Orthosis Type		Fitting Date			
Assistive Devices (cane, crutches, walker, etc.)		Brief History			
Years of orthosis use					

Part 1: Please rate the following items for the Ottawalk-Speed stance control-knee-ankle-foot-orthosis (OWS) by placing a mark (×) in the appropriate box.

	Very Difficult	**Difficult**	**Neutral**	**Easy**	**Very Easy**
Putting on the OWS					
Applying the straps					
Removing the OWS					
Putting on your shoe					
Balance when using the OWS					
Sitting when using the OWS					
Stability during level walking					
Stability when walking on uneven ground					

Part 2: Please rate the following items by circling the appropriate box.

Attitude toward wearing the OWS	**Enthusiastic**	**Accepting**	**Neutral**	**Disinterested**	**Rejecting**

Is the OWS comfortable	**All the Time**	**Most of the Time**	**About Half the Time**	**Hardly Ever**	**Never**

Walking ability when using the OWS SCKAFO, compared to walking with current orthosis	**Much Improved**	**Somewhat Improved**	**Same**	**Somewhat Worse**	**Much Worse**

OWS appearance	**Very Good**	**Good**	**Acceptable**	**Poor**	**Very Poor**

Effort when using OWS, compared to current orthosis	**Much Less**	**Somewhat Less**	**Same**	**Somewhat More**	**Much More**

The weight of the OWS is	**Very Light**	**Light**	**Not Light or Heavy**	**Heavy**	**Very Heavy**

OWS support	**Very Secure**	**Somewhat Secure**	**Same as Without Orthosis**	**Somewhat Insecure**	**Very Insecure**

Donning and doffing	**Very Easy**	**Easy**	**Some Trouble**	**Much Trouble**	**Cannot Do Unassisted**

	Yes	**No**
Does the OWS make you perspire?		
Does the OWS rub your skin?		

What do you like best about the OWS?	
What do you like least about the OWS?	
Additional Comments	

Appendix I: Knee Flexion at TL FS and REC1FS for Stumble Development Participants

Table 16: Knee flexion angle (degrees) for stumble development participants.

	Trial	Trip Leg	TL	REC1FS	TL Max 1	Recovery 1 Max 1
Participant 1	1	R	4.71	18.80	18.51	29.40
	2	L	6.49	11.16	20.37	27.96
	3	R	7.25	16.69	19.96	30.01
	4	L	7.48	17.16	20.83	30.53
	5	L	7.35	16.60	21.70	33.01
	6	R	8.22	20.30	22.73	25.80
	7	R	8.50	22.94	24.17	30.76

	Trial	Trip Leg	TL	REC1FS	TL Max 1	Recovery 1 Max 1
Participant 2	1	R	5.71	27.82	17.11	37.25
	2	L	5.15	22.93	17.13	45.79
	3	R	8.79	27.81	18.05	37.02
	4	L	9.65	31.15	18.66	47.35
	5	L	9.66	31.77	17.91	40.94
	6	R	9.30	27.57	18.43	37.17
	7	L	11.32	30.87	20.45	43.06
	8	R	11.88	28.36	19.61	39.38
	9	R	10.46	28.93	18.25	39.90
	10	L	10.09	37.68	17.13	42.28

	Trial	Trip Leg	TL	REC1FS	TL Max 1	Recovery 1 Max 1
Participant 3	1	R	3.89	34.91	18.29	49.89
	2	L	4.12	30.26	22.08	41.43
	3	R	4.96	26.42	21.97	51.17
	4	L	5.51	30.31	22.68	47.06
	5	R	5.02	15.57	21.67	45.62
	6	L	3.78	29.72	21.43	41.64
	7	L	4.74	26.56	22.42	45.14
	8	R	4.04	20.38	22.90	45.48
	9	R	3.76	20.59	22.63	41.99
	10	L	2.92	25.86	21.51	43.31

	Trial	Trip Leg	TL	REC1FS	TL Max 1	Recovery 1 Max 1
Participant 4	1	R	6.76	23.04	20.28	30.67
	2	L	9.36	27.24	21.42	34.35
	3	L	9.88	22.60	20.84	32.64
	4	R	9.26	24.09	22.75	30.84
	5	L	12.42	25.24	22.18	25.70
	6	R	12.00	19.98	24.17	28.02
	7	L	12.54	27.35	25.80	35.48
	8	R	10.62	20.01	26.21	35.22
	9	R	12.41	26.89	25.75	31.20
	10	L	13.34	32.11	27.44	36.66

	Trial	Trip Leg	TL	REC1FS	TL Max 1	Recovery 1 Max 1
Participant 5	1	L	0.84	32.94	12.96	120.30
	2	R	3.62	25.60	10.05	42.84
	3	L	-2.21	21.71	11.63	67.14
	4	R	-1.69	17.94	10.61	115.57
	5	R	2.33	19.12	8.31	52.44
	6	L	0.52	33.69	7.34	55.99
	7	L	-0.77		8.20	
	8	R	-1.82	45.16	8.66	112.29
	9	R	-1.53	22.68	9.11	94.19
	10	L	-0.37	70.77	10.19	115.83

Appendix J: Ankle Joint Angle Curves for Stumble Development Participants

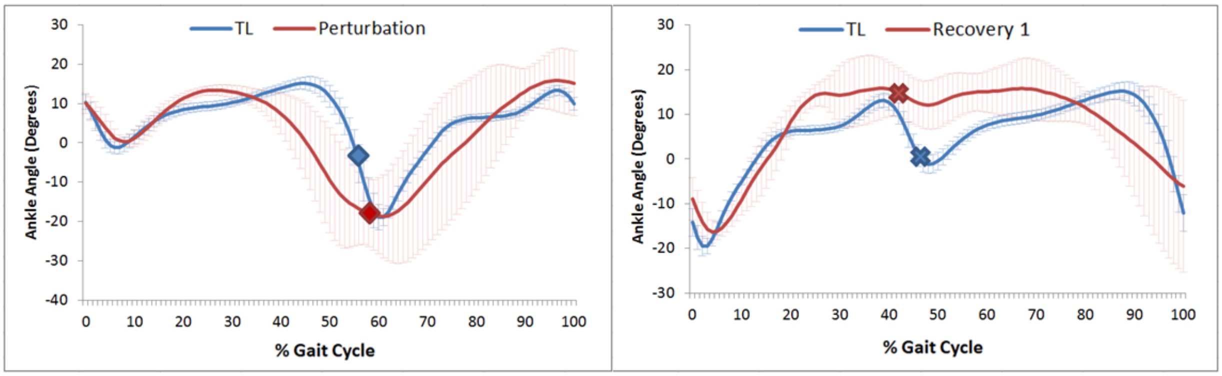

Figure 20: Ankle dorsiflexion (+) and plantar flexion (-), normalized to 100% of the gait cycle, for stumble development participant 1: FS-FS for TL and perturbation strides (left) and FO-FO for TL and recovery 1 strides (right).

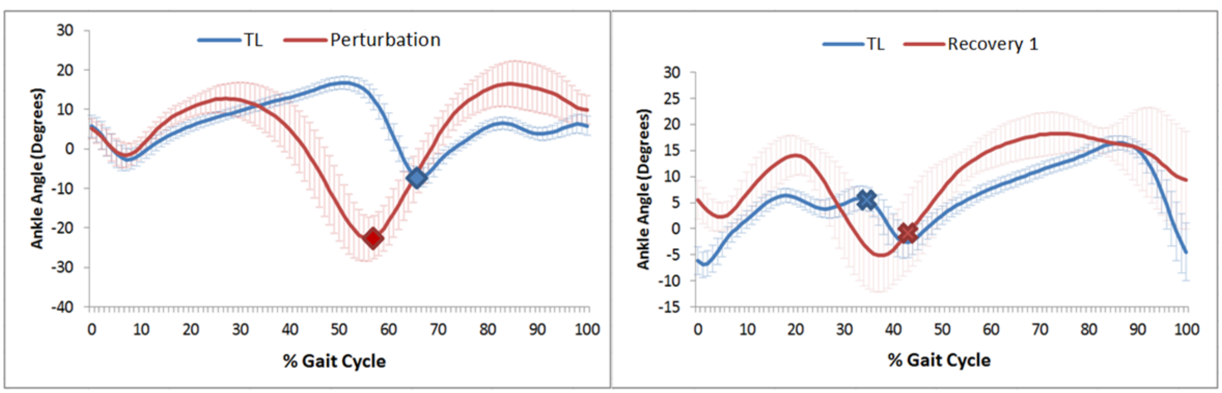

Figure 21: Ankle dorsiflexion (+) and plantar flexion (-), normalized to 100% of the gait cycle, for stumble development participant 2: FS-FS for TL and perturbation strides (left) and FO-FO for TL and recovery 1 strides (right).

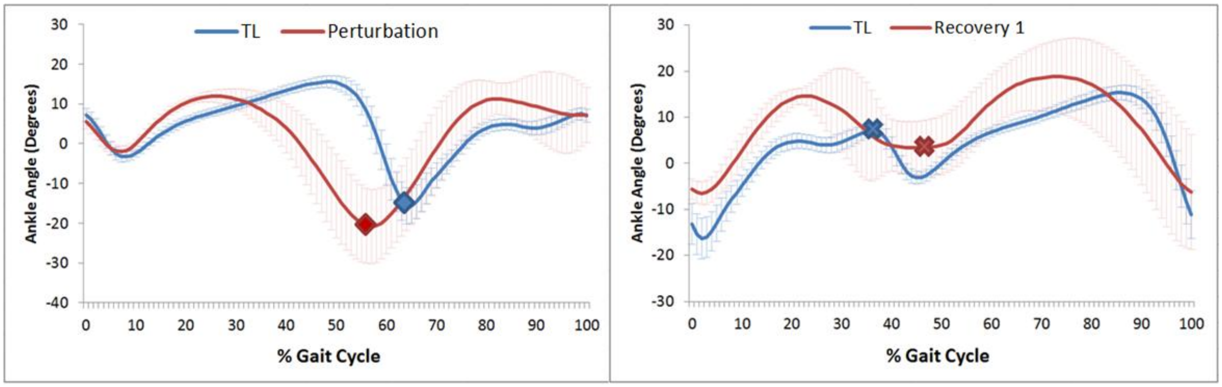

Figure 22: Ankle dorsiflexion (+) and plantar flexion (-), normalized to 100% of the gait cycle, for stumble development participant 3: FS-FS for TL and perturbation strides (left) and FO-FO for TL and recovery 1 strides (right).

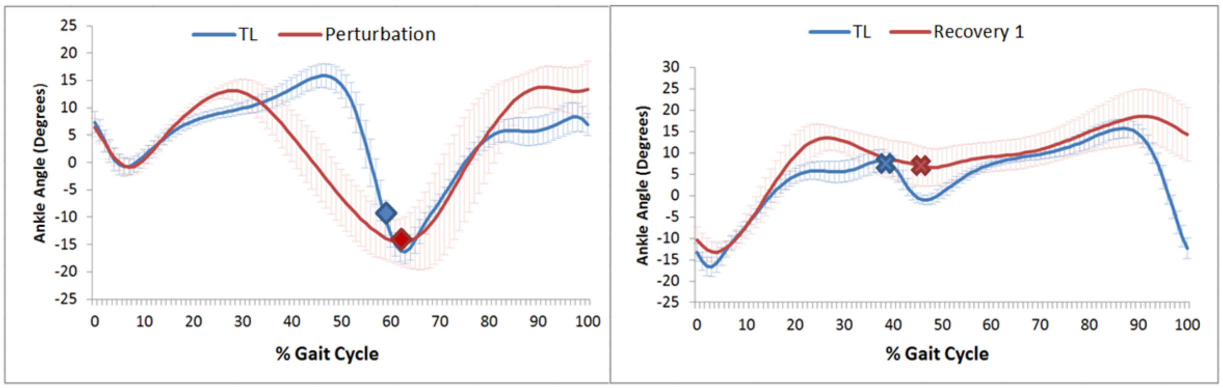

Figure 23: Ankle dorsiflexion (+) and plantar flexion (-), normalized to 100% of the gait cycle, for stumble development participant 4: FS-FS for TL and perturbation strides (left) and FO-FO for TL and recovery 1 strides (right).

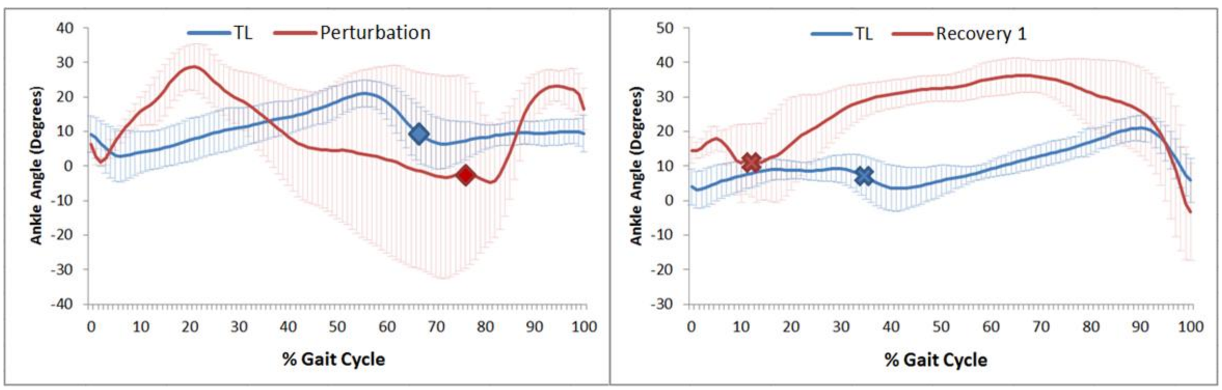

Figure 24: Ankle dorsiflexion (+) and plantar flexion (-), normalized to 100% of the gait cycle, for stumble development participant 5: FS-FS for TL and perturbation strides (left) and FO-FO for TL and recovery 1 strides (right).

Appendix K: Knee Joint Angle for Stumble Development Participants

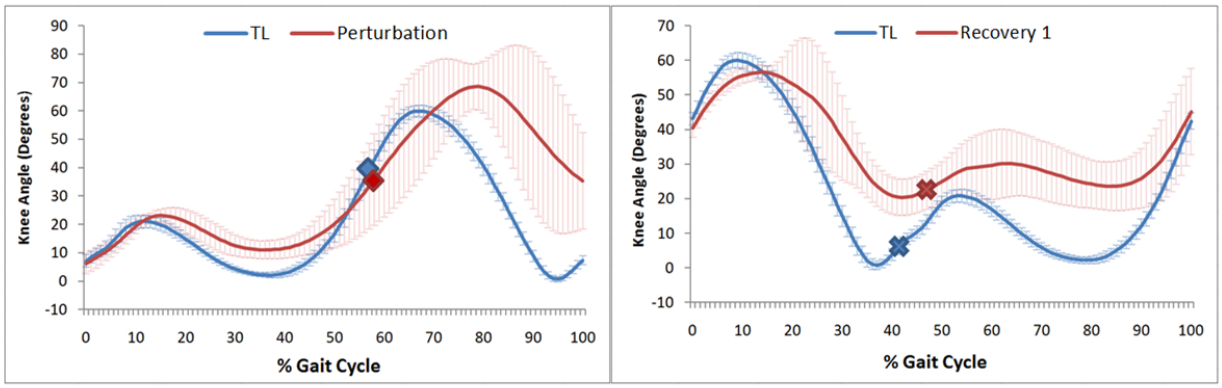

Figure 25: Knee flexion (+) and extension (-), normalized to 100% of the gait cycle, for stumble development participant 1: FS-FS for TL and perturbation strides (left) and FO-FO for TL and recovery 1 strides (right).

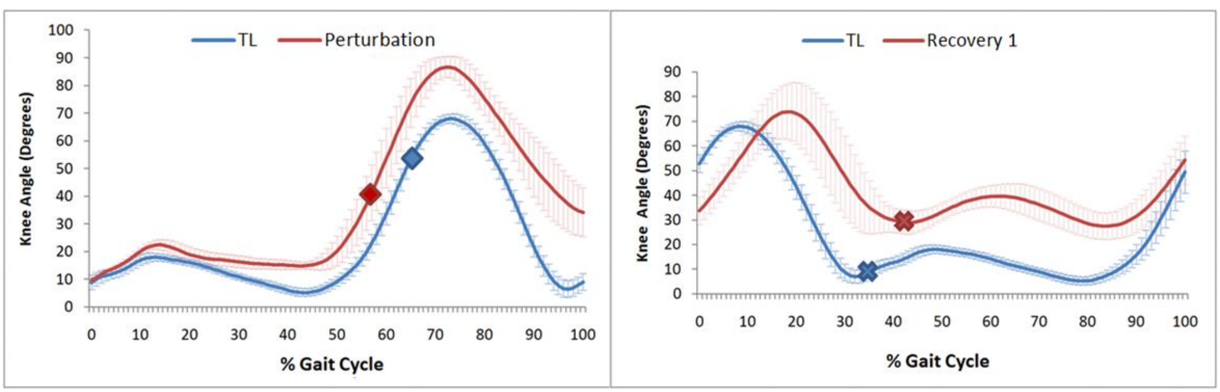

Figure 26: Knee flexion (+) and extension (-), normalized to 100% of the gait cycle, for stumble development participant 2: FS-FS for TL and perturbation strides (left) and FO-FO for TL and recovery 1 strides (right).

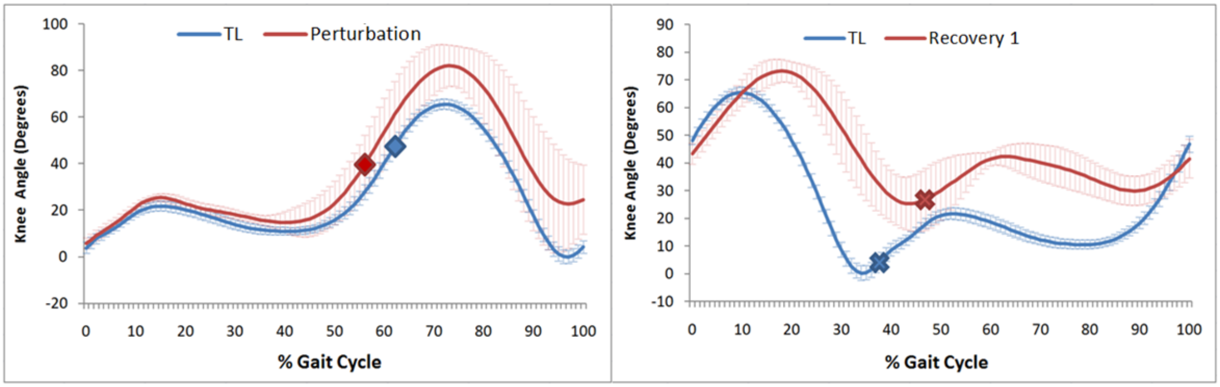

Figure 27: Knee flexion (+) and extension (-), normalized to 100% of the gait cycle, for stumble development participant 3: FS-FS for TL and perturbation strides (left) and FO-FO for TL and recovery 1 strides (right).

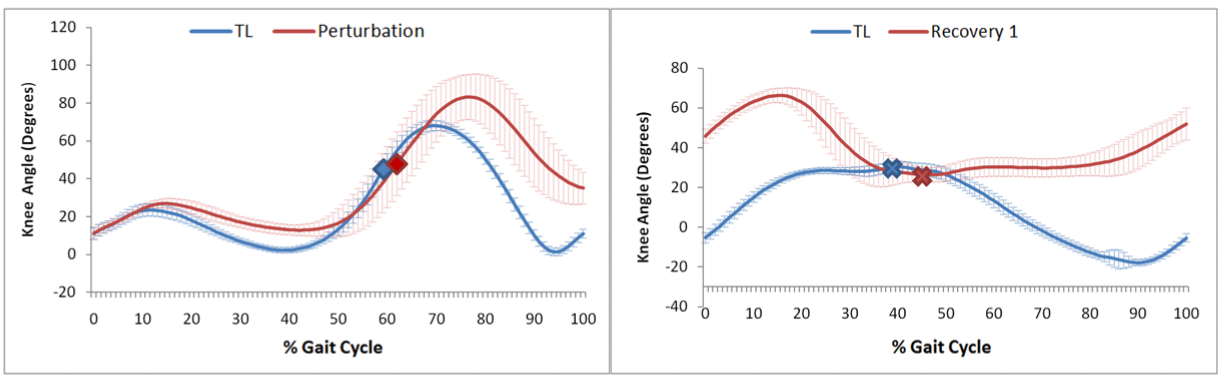

Figure 28: Knee flexion (+) and extension (-), normalized to 100% of the gait cycle, for stumble development participant 4: FS-FS for TL and perturbation strides (left) and FO-FO for TL and recovery 1 strides (right).

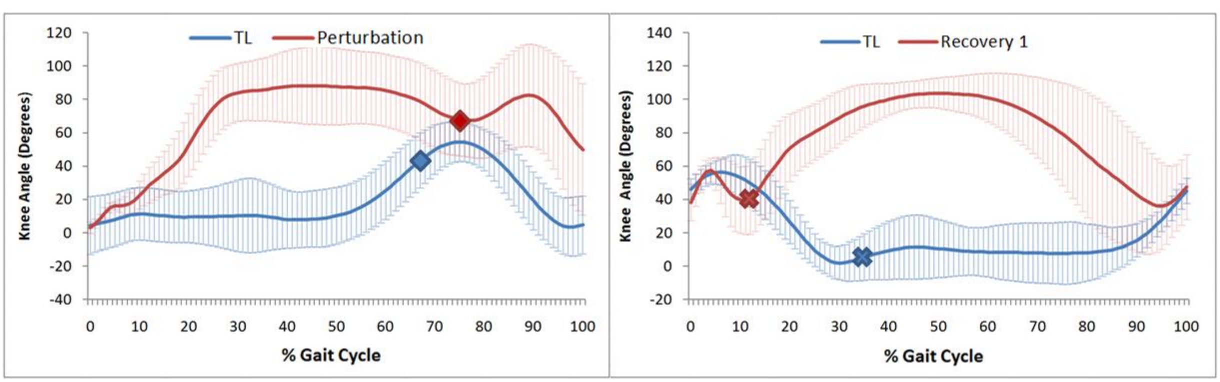

Figure 29: Knee flexion (+) and extension (-), normalized to 100% of the gait cycle, for stumble development participant 5: FS-FS for TL and perturbation strides (left) and FO-FO for TL and recovery 1 strides (right).

Appendix L: Hip Joint Angle for Stumble Development Participants

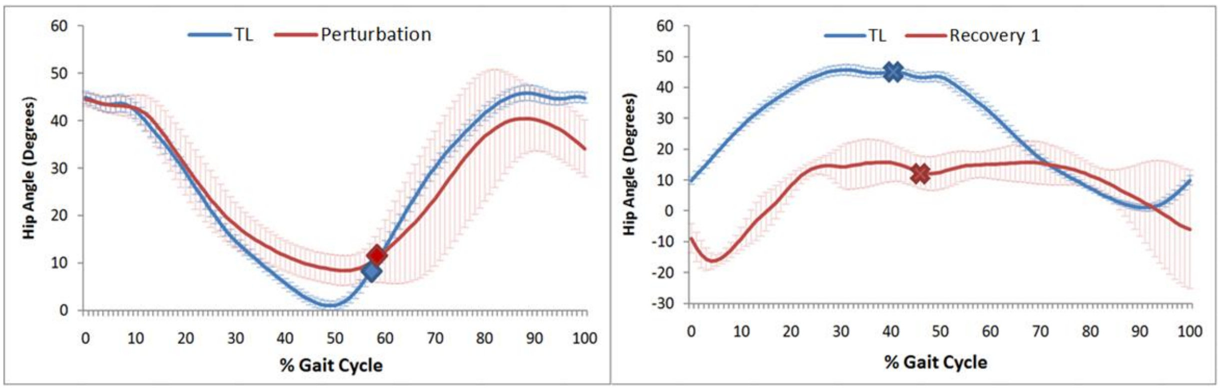

Figure 30: Hip flexion (+) and extension (-), normalized to 100% of the gait cycle, for stumble development participant 1: FS-FS for TL and perturbation strides (left) and FO-FO for TL and recovery 1 strides (right).

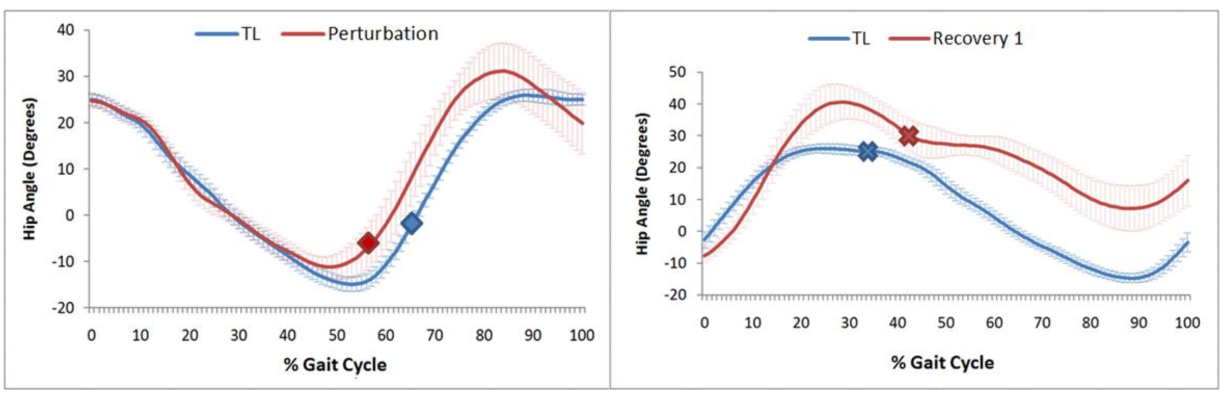

Figure 31: Hip flexion (+) and extension (-), normalized to 100% of the gait cycle, for stumble development participant 2: FS-FS for TL and perturbation strides (left) and FO-FO for TL and recovery 1 strides (right).

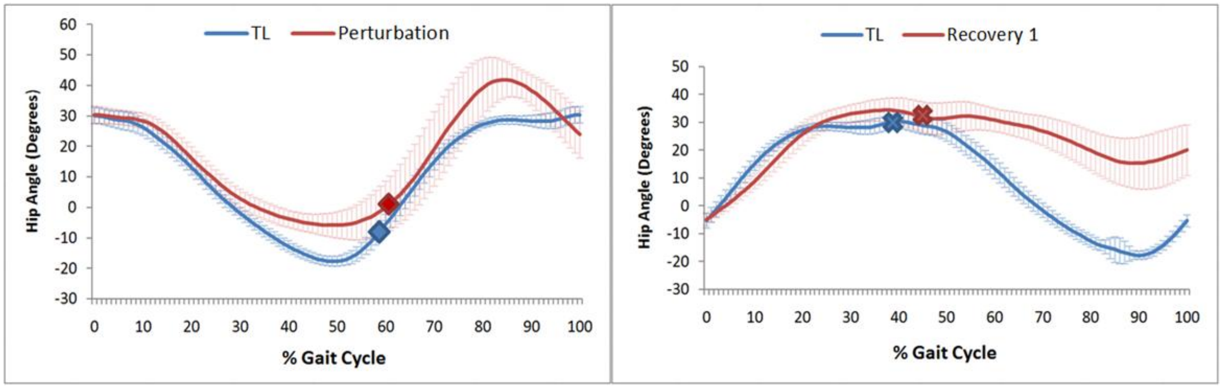

Figure 32 Hip flexion (+) and extension (-), normalized to 100% of the gait cycle, for stumble development participant 3: FS-FS for TL and perturbation strides (left) and FO-FO for TL and recovery 1 strides (right).

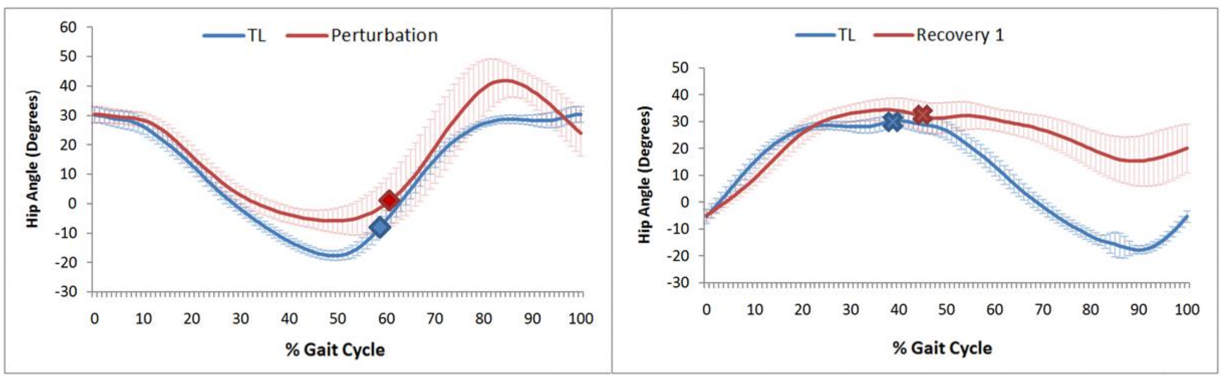

Figure 33: Hip flexion (+) and extension (-), normalized to 100% of the gait cycle, for stumble development participant 4: FS-FS for TL and perturbation strides (left) and FO-FO for TL and recovery 1 strides (right).

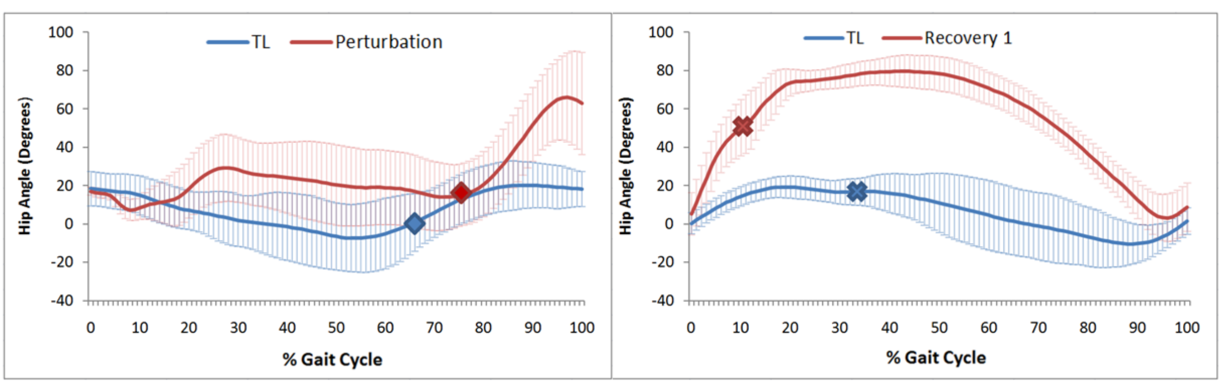

Figure 34: Hip flexion (+) and extension (-), normalized to 100% of the gait cycle, for stumble development participant 5: FS-FS for TL and perturbation strides (left) and FO-FO for TL and recovery 1 strides (right).

Appendix M: Pelvis Angle for Stumble Development Participants

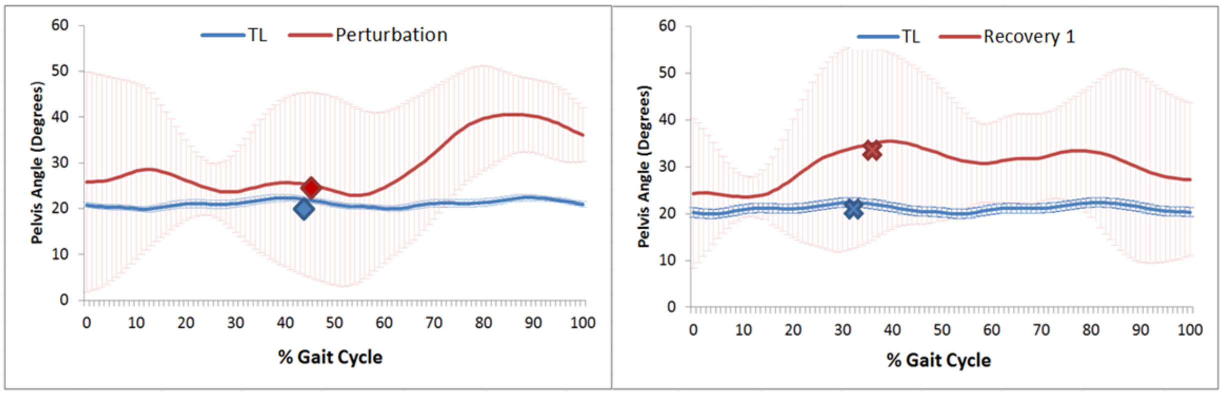

Figure 35: Pelvic anterior tilt (+) and posterior tilt (-), normalized to 100% of the gait cycle, for stumble development participant 1: FS-FS for TL and perturbation strides (left) and FO-FO for TL and recovery 1 strides (right).

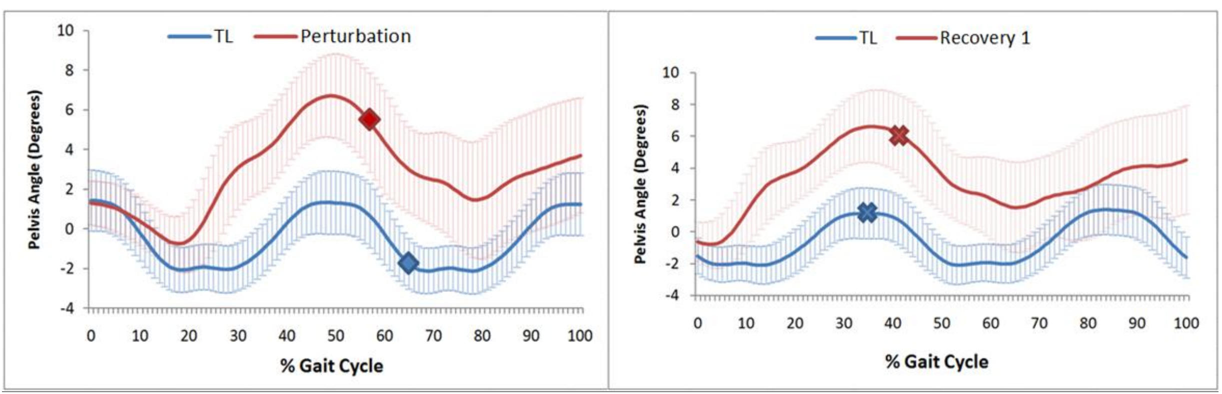

Figure 36: Pelvic anterior tilt (+) and posterior tilt (-), normalized to 100% of the gait cycle, for stumble development participant 2: FS-FS for TL and perturbation strides (left) and FO-FO for TL and recovery 1 strides (right).

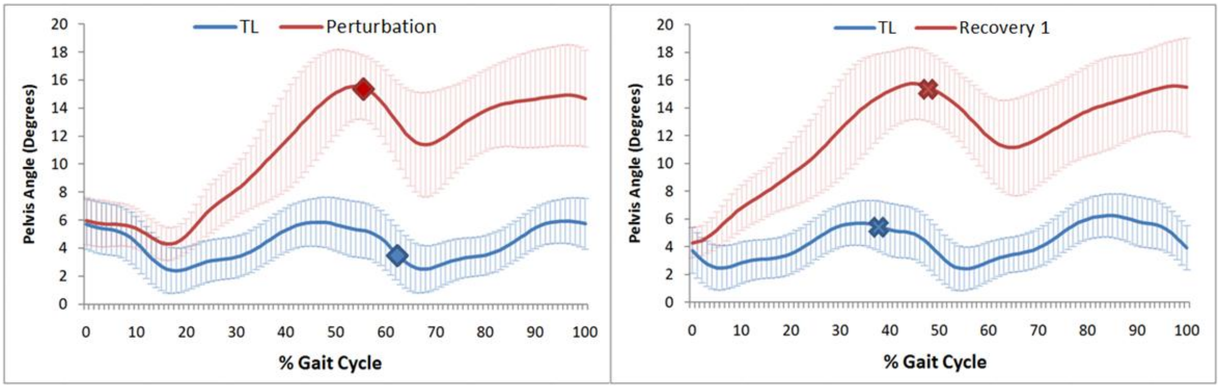

Figure 37: Pelvic anterior tilt (+) and posterior tilt (-), normalized to 100% of the gait cycle, for stumble development participant 3: FS-FS for TL and perturbation strides (left) and FO-FO for TL and recovery 1 strides (right).

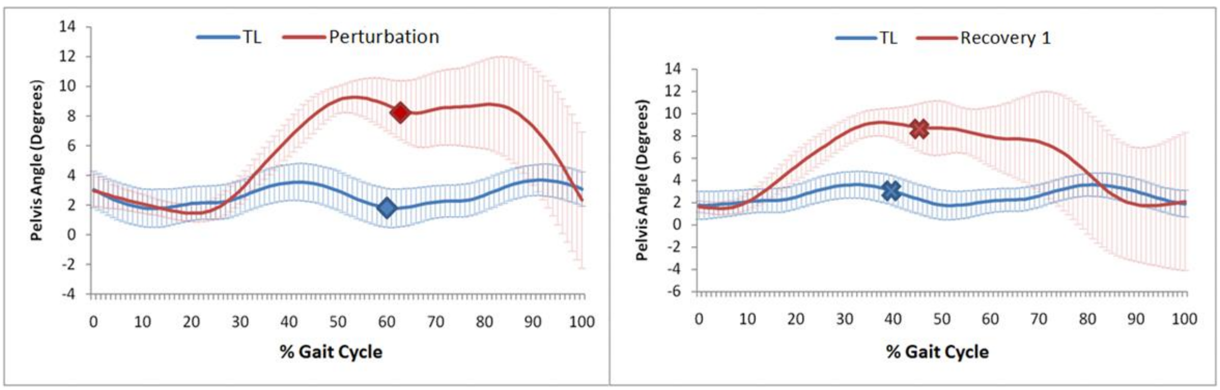

Figure 38: Pelvic anterior tilt (+) and posterior tilt (-), normalized to 100% of the gait cycle, for stumble development participant 4: FS-FS for TL and perturbation strides (left) and FO-FO for TL and recovery 1 strides (right).

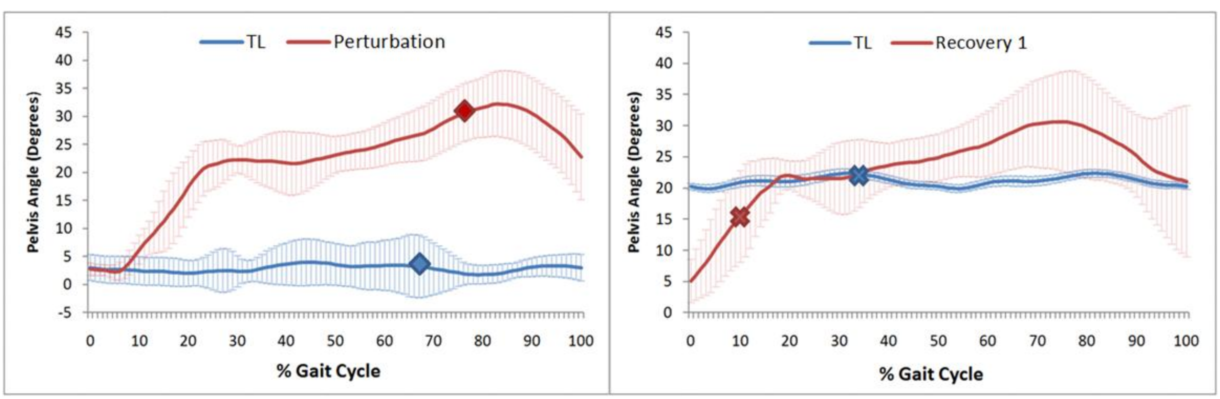

Figure 39: Pelvic anterior tilt (+) and posterior tilt (-), normalized to 100% of the gait cycle, for stumble development participant 5: FS-FS for TL and perturbation strides (left) and FO-FO for TL and recovery 1 strides (right).

Appendix N: Knee Joint Angular Velocity for Stumble Development Participants

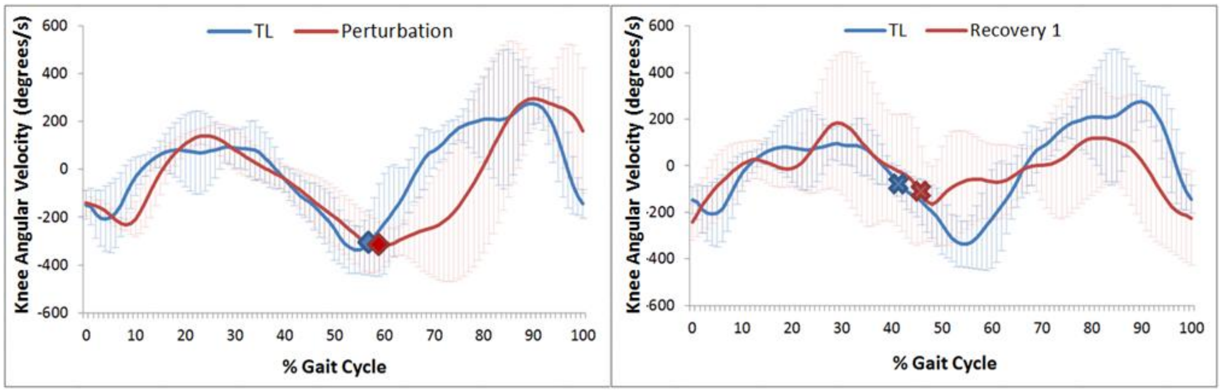

Figure 40: Knee extension (+) and flexion (-) angular velocity, normalized to 100% of the gait cycle, for stumble development participant 1: FS-FS for TL and perturbation strides (left) and FO-FO for TL and recovery 1 strides (right).

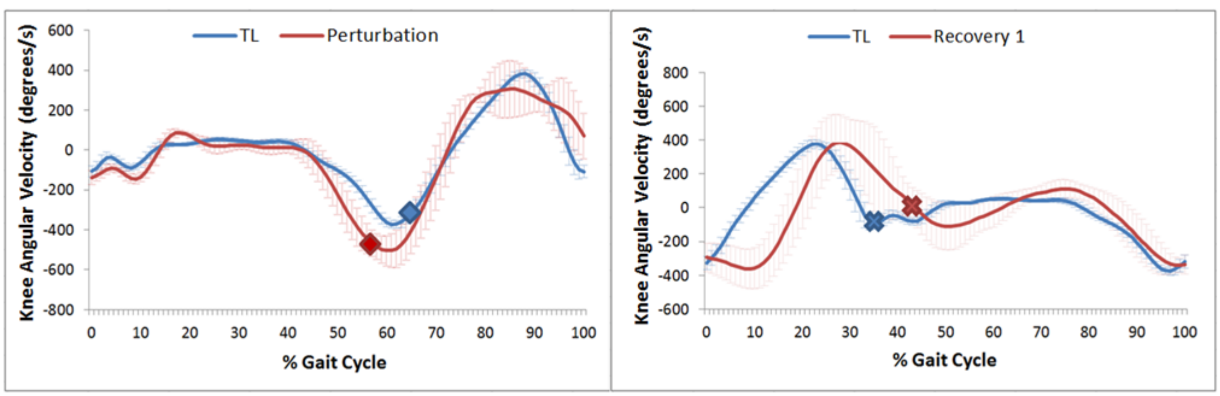

Figure 41: Knee extension (+) and flexion (-) angular velocity, normalized to 100% of the gait cycle, for stumble development participant 2: FS-FS for TL and perturbation strides (left) and FO-FO for TL and recovery 1 strides (right).

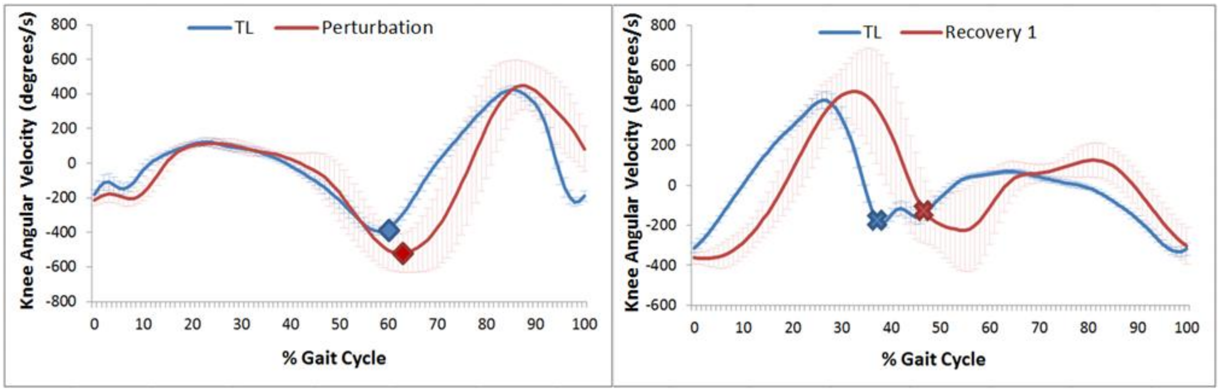

Figure 42: Knee extension (+) and flexion (-) angular velocity, normalized to 100% of the gait cycle, for stumble development participant 3: FS-FS for TL and perturbation strides (left) and FO-FO for TL and recovery 1 strides (right).

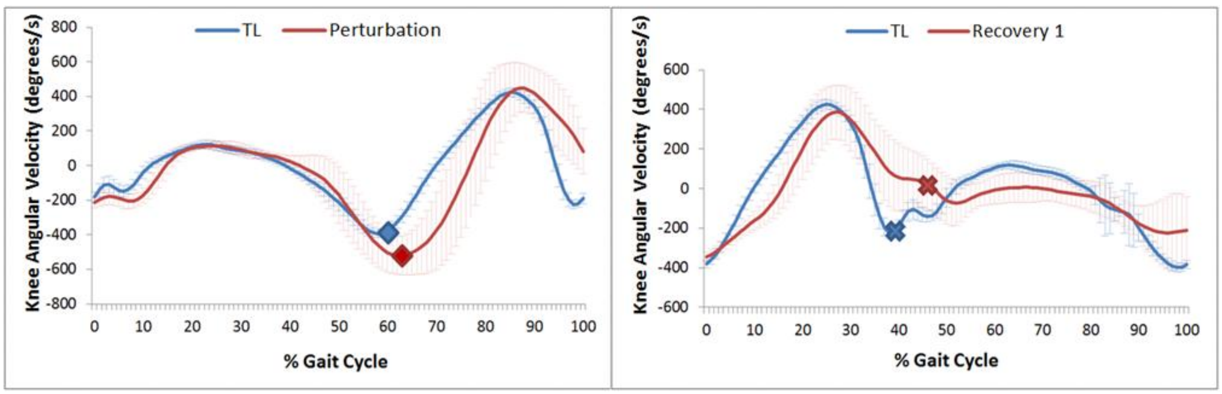

Figure 43: Knee extension (+) and flexion (-) angular velocity, normalized to 100% of the gait cycle, for stumble development participant 4: FS-FS for TL and perturbation strides (left) and FO-FO for TL and recovery 1 strides (right).

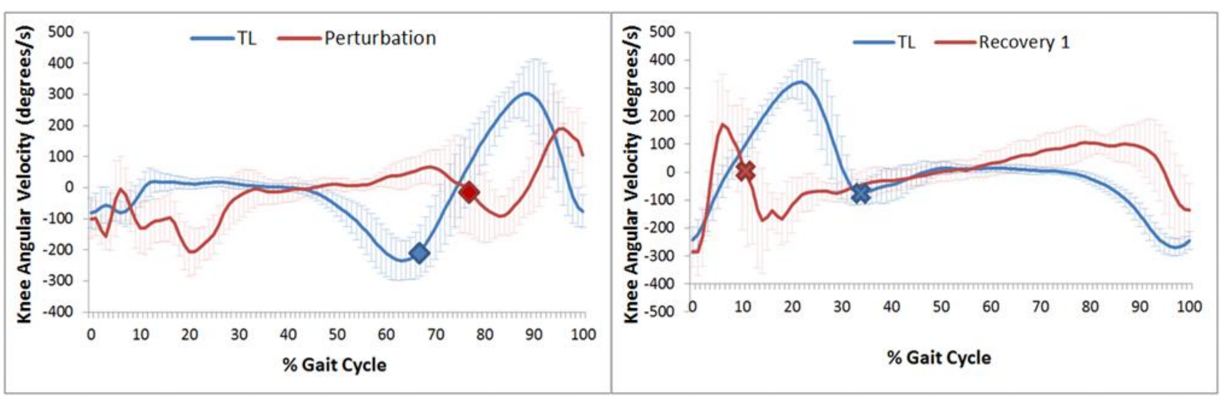

Figure 44: Knee extension (+) and flexion (-) angular velocity, normalized to 100% of the gait cycle, for stumble development participant 5: FS-FS for TL and perturbation strides (left) and FO-FO for TL and recovery 1 strides (right).

Table 17: Hopping strategy angular velocity averages (degrees/s) and standard deviations at FS and FO for TL and perturbed gait events. Participants 1-4.

	TL FS	PERTFS	REC1FS	REC2FS
Ankle	-107.04 ± 60.64	-146.14 ± 63.63	44.38 ± 198.87	10.03 ± 85.03
Knee	-161.64 ± 40.30	-181.30 ± 49.36	-65.44 ± 118.77	55.90 ± 110.70
Hip	-2.84 ± 20.31	-4.89 ± 21.58	-83.62 ± 44.13	-124.36 ± 60.56
	TL FO	**PERTFO**	**REC1FO**	**REC2FO**
Ankle	-259.34 ± 122.32	-59.20 ± 85.01	-73.90 ± 45.76	-121.63 ± 81.61
Knee	-339.11 ± 32.85	-450.96 ± 99.20	-306.65 ± 72.74	-352.07 ± 63.06
Hip	166.19 ± 23.88	64.15 ± 226.54	121.88 ± 40.95	150.97 ± 44.05

* Ankle dorsiflexion (+) and plantar flexion (-), knee extension (+) and flexion (-), and hip flexion (+) and extension (-)

Table 18: Stopping strategy angular velocity averages (degrees/s) and standard deviations at FS and FO for TL and perturbed gait events. Participant 5.

	TL FS	PERTFS	REC1FS	REC2FS
Ankle	-75.40 ± 49.69	-161.47 ± 38.43	-184.10 ± 121.83	-263.56 ± 151.17
Knee	-83.48 ± 33.16	-116.23 ± 43.70	-51.99 ± 31.06	148.31 ± 72.88
Hip	-0.55 ± 19.43	-22.20 ± 38.79	78.07 ± 44.73	-71.62 ± 72.85
	TL FO	**PERTFO**	**REC1FO**	**REC2FO**
Ankle	-111.64 ± 89.94	51.88 ± 68.84	-21.71 ± 157.53	-26.08 ± 68.96
Knee	-239.95 ± 36.72	-32.96 ± 80.04	-92.51 ± 85.33	-224.40 ± 65.50
Hip	129.17 ± 23.79	22.47 ± 77.90	83.50 ± 35.53	114.55 ± 34.78

* Ankle dorsiflexion (+) and plantar flexion (-), knee extension (+) and flexion (-), and hip flexion (+) and extension (-)

Appendix O: Ankle Joint Angles for OWS Participants

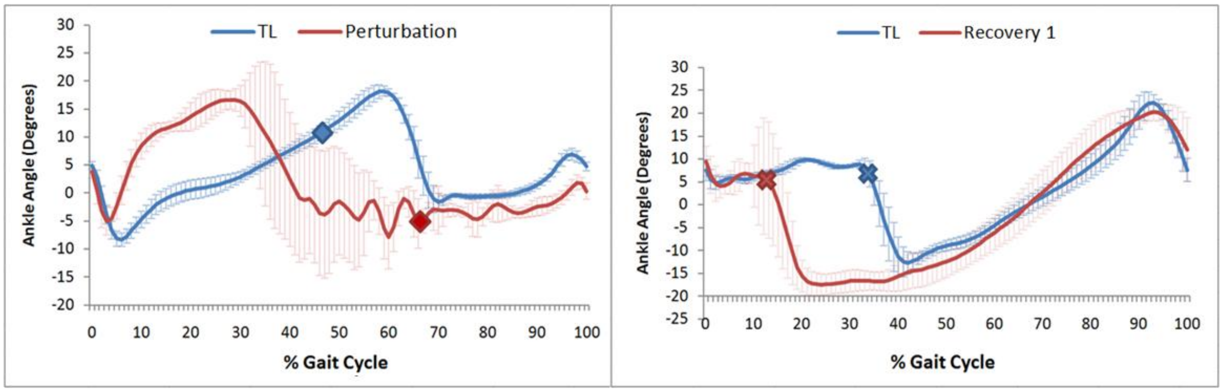

Figure 45: Ankle dorsiflexion (+) and plantar flexion (-), normalized to 100% of the gait cycle, for OWS participant 1: FS-FS for TL and perturbation strides (left) and FO-FO for TL and recovery 1 strides (right).

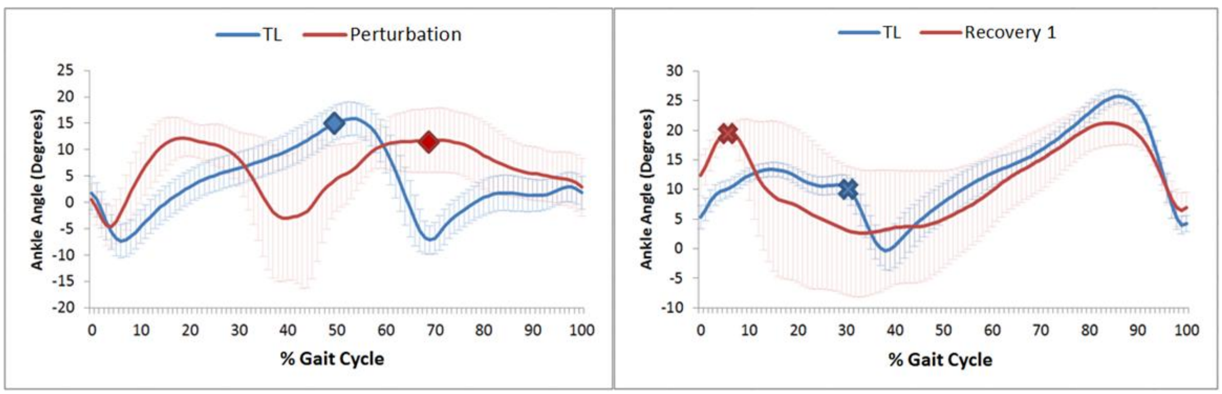

Figure 46: Ankle dorsiflexion (+) and plantar flexion (-), normalized to 100% of the gait cycle, for OWS participant 2: FS-FS for TL and perturbation strides (left) and FO-FO for TL and recovery 1 strides (right).

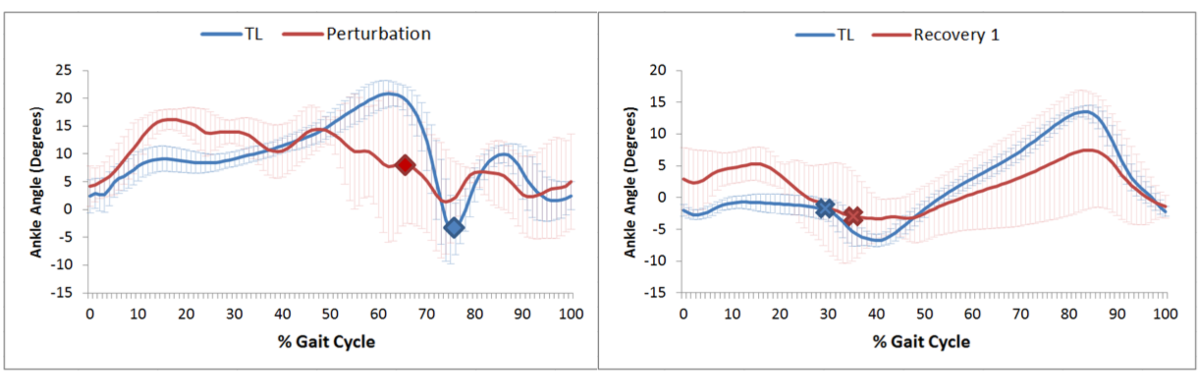

Figure 47 Ankle dorsiflexion (+) and plantar flexion (-), normalized to 100% of the gait cycle, for OWS participant 3: FS-FS for TL and perturbation strides (left) and FO-FO for TL and recovery 1 strides (right).

Appendix P: Hip Joint Angles for OWS Participants

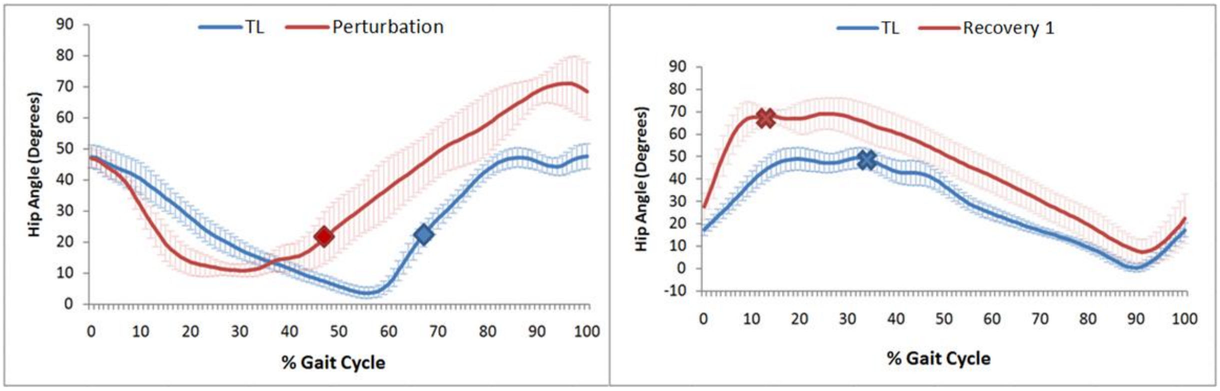

Figure 48: Hip flexion (+) and extension (-), normalized to 100% of the gait cycle, for OWS participant 1: FS-FS for TL and perturbation strides (left) and FO-FO for TL and recovery 1 strides (right).

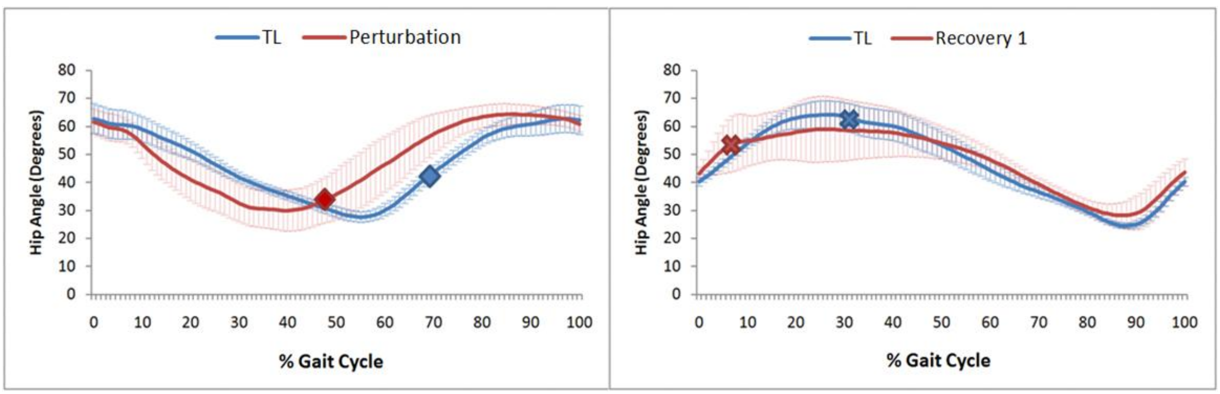

Figure 49: Hip flexion (+) and extension (-), normalized to 100% of the gait cycle, for OWS participant 2: FS-FS for TL and perturbation strides (left) and FO-FO for TL and recovery 1 strides (right).

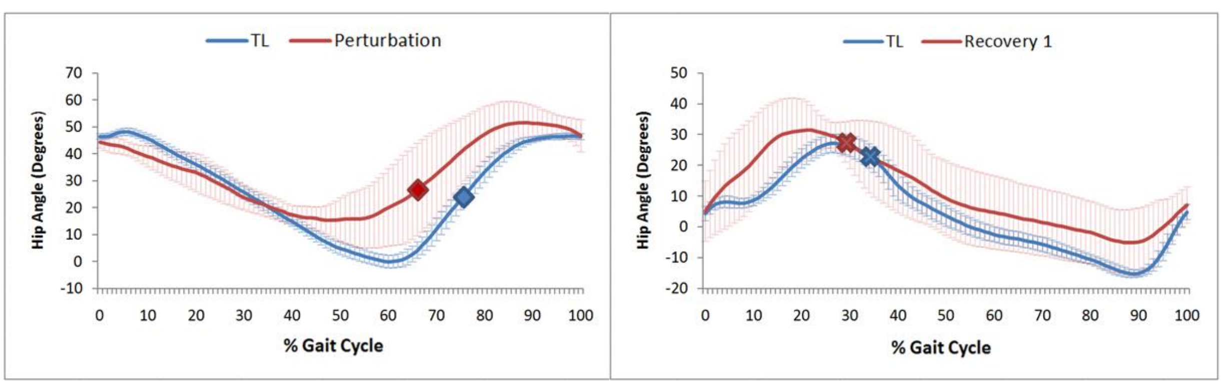

Figure 50: Hip flexion (+) and extension (-), normalized to 100% of the gait cycle, for OWS participant 3: FS-FS for TL and perturbation strides (left) and FO-FO for TL and recovery 1 strides (right).

Appendix Q: Pelvis Angle for OWS Participants

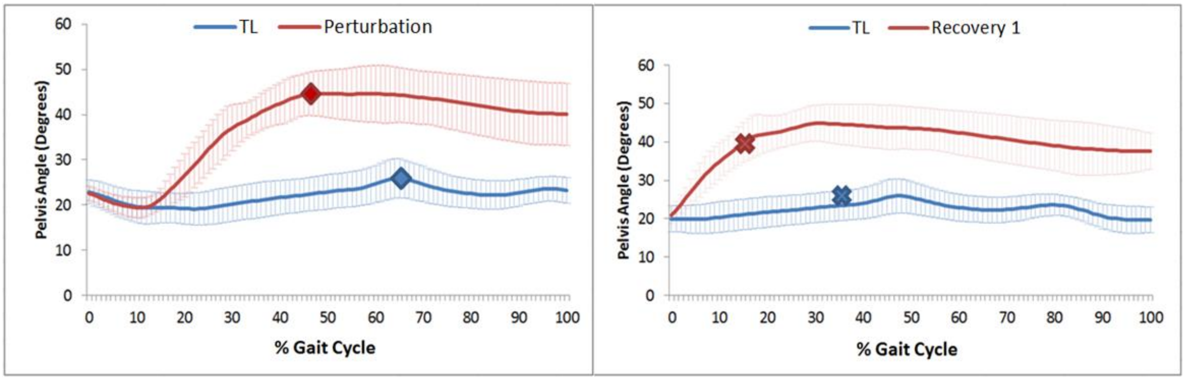

Figure 51: Pelvic anterior tilt (+) and posterior tilt (-), normalized to 100% of the gait cycle, for OWS participant 1: FS-FS for TL and perturbation strides (left) and FO-FO for TL and recovery 1 strides (right).

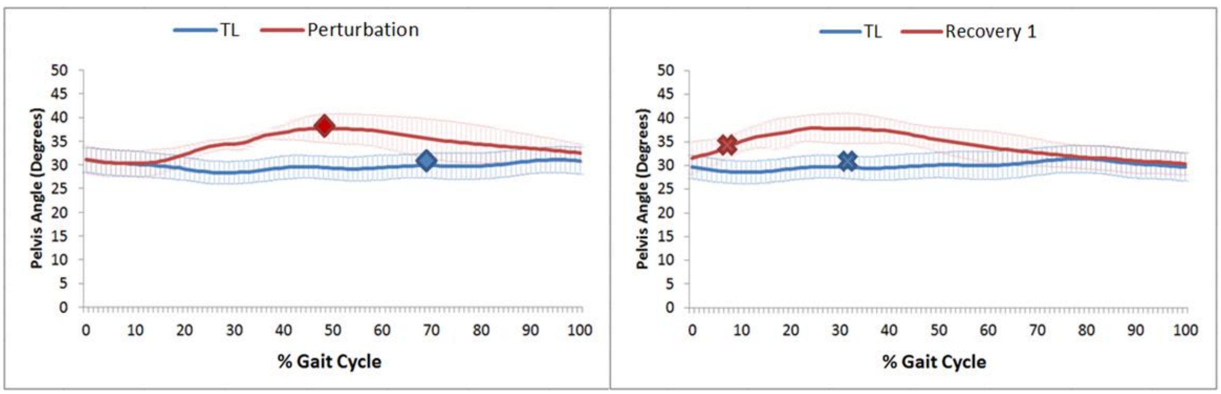

Figure 52: Pelvic anterior tilt (+) and posterior tilt (-), normalized to 100% of the gait cycle, for OWS participant 2: FS-FS for TL and perturbation strides (left) and FO-FO for TL and recovery 1 strides (right).

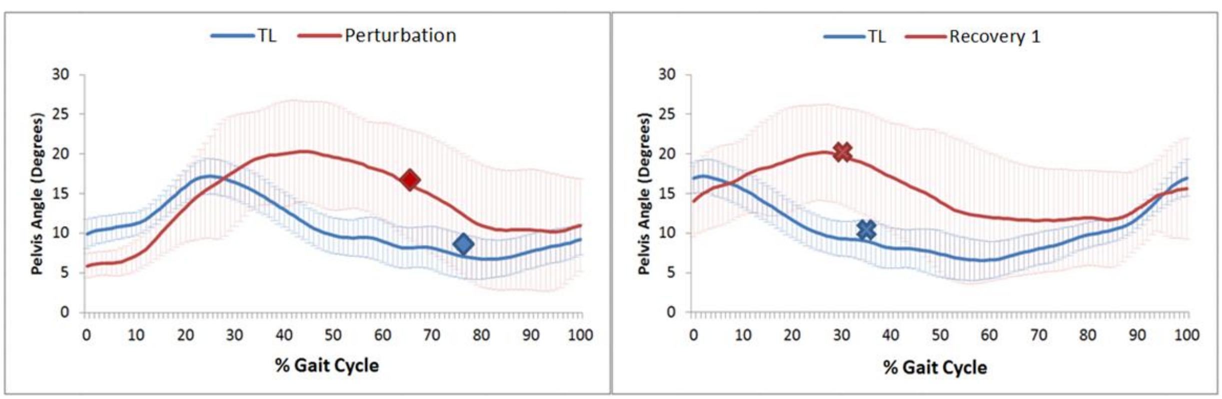

Figure 53: Pelvic anterior tilt (+) and posterior tilt (-), normalized to 100% of the gait cycle, for OWS participant 3: FS-FS for TL and perturbation strides (left) and FO-FO for TL and recovery 1 strides (right).

Appendix R: Knee Angular Velocity - OWS Participants

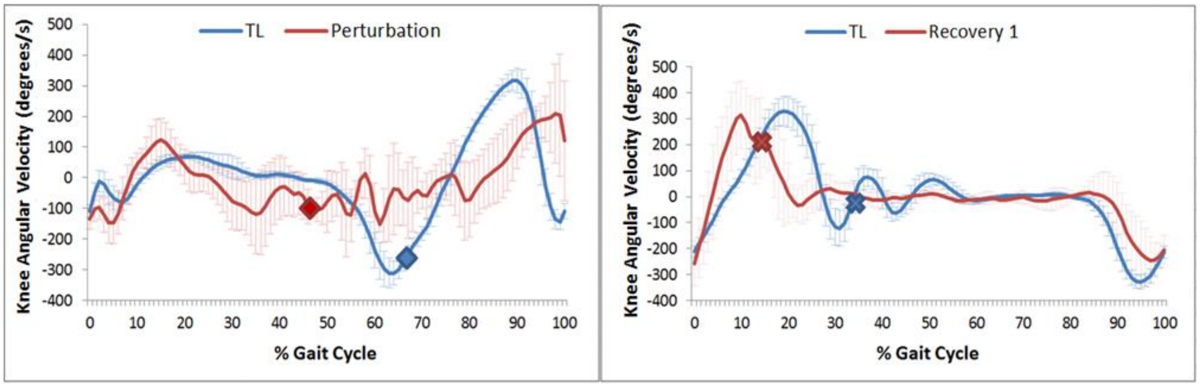

Figure 54: Knee extension (+) and flexion (-) angular velocity, normalized to 100% of the gait cycle, for OWS participant 1: FS-FS for TL and perturbation strides (left) and FO-FO for TL and recovery 1 strides (right).

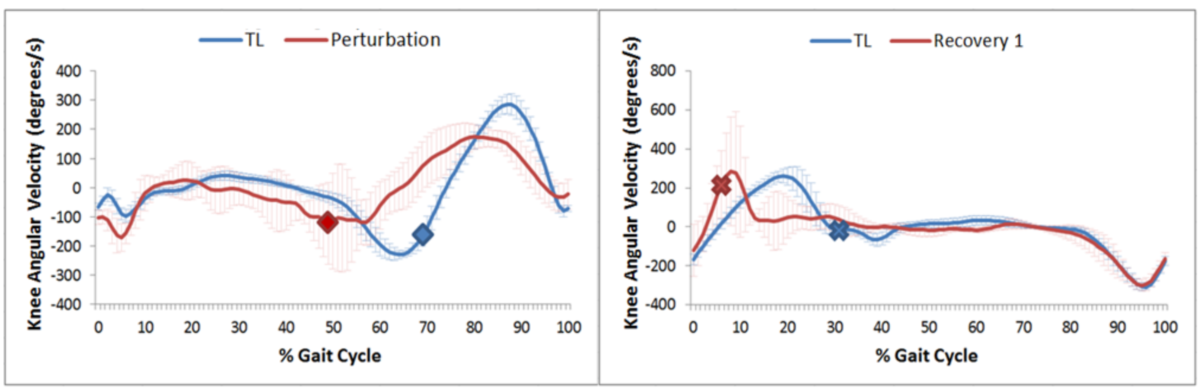

Figure 55: Knee extension (+) and flexion (-) angular velocity, normalized to 100% of the gait cycle, for OWS participant 2: FS-FS for TL and perturbation strides (left) and FO-FO for TL and recovery 1 strides (right).

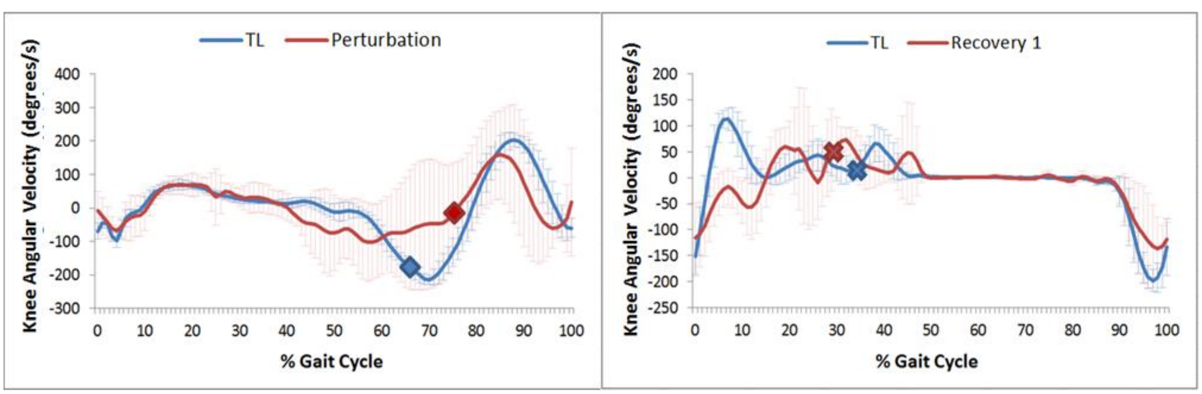

Figure 56: Knee extension (+) and flexion (-) angular velocity, normalized to 100% of the gait cycle, for OWS participant 3: FS-FS for TL and perturbation strides (left) and FO-FO for TL and recovery 1 strides (right).

Table 19: OWS participant 1 angular velocity averages (degrees/s) and standard deviations at FS and FO for TL and perturbed gait events: ankle dorsiflexion (+) and plantar flexion (-), knee extension (+) and flexion (-), and hip flexion (+) and extension (-).

	Non-Braced			**Braced**	
	TL Left FS	**PERTFS**	**REC2FS**	**TL Right FS**	**REC1FS**
Ankle	-96.21 ± 46.93	-157.94 ± 67.40	-73.92 ± 94.28	-86.07 ± 19.83	-164.24 ± 53.32
Knee	-109.25 ± 19.94	-133.24 ± 34.09	121.48 ± 195.12	-36.83 ± 53.93	231.44 ± 92.14
Hip	0.92 ± 15.07	-8.84 ± 31.44	-56.14 ± 64.73	-26.83 ± 24.63	-37.96 ± 24.46
	Non-Braced			**Braced**	
	TL Left FO	**PERTFO**	**REC2FO**	**TL Right FO**	**REC1FO**
Ankle	-196.81 ± 27.90	-28.76 ± 108.14	-120.97 ± 98.14	-140.35 ± 32.80	-75.97 ± 53.34
Knee	-269.25 ± 13.66	-90.17 ± 89.20	-228.52 ± 62.70	-211.68 ± 15.69	-203.84 ± 56.34
Hip	117.62 ± 7.08	52.90 ± 36.19	107.07 ± 9.58	129.38 ± 16.08	124.03 ± 21.93

Table 20: OWS participant 2 angular velocity averages (degrees /s) and standard deviations at FS and FO for TL and perturbed gait events: ankle dorsiflexion (+) and plantar flexion (-), knee extension (+) and flexion (-), and hip flexion (+) and extension (-).

	Non-Braced			**Braced**	
	TL Left FS	**PERTFS**	**REC2FS**	**TL Right FS**	**REC1FS**
Ankle	-74.50 ± 25.97	-115.57 ± 56.78	-66.47 ± 58.66	-90.80 ± 36.75	-124.45 ± 152.04
Knee	-67.34 ± 8.83	-102.37 ± 25.53	-22.63 ± 50.79	-21.93 ± 5.84	101.16 ± 139.03
Hip	-33.63 ± 9.25	-44.28 ± 6.72	-59.15 ± 28.96	-42.95 ± 6.52	28.11 ± 22.99
	Non-Braced			**Braced**	
	TL Left FO	**PERTFO**	**REC2FO**	**TL Right FO**	**REC1FO**
Ankle	-19.69 ± 44.18	177.90 ± 170.21	30.59 ± 36.78	61.22 ± 27.89	62.40 ± 48.96
Knee	-183.76 ± 21.51	-191.93 ± 82.50	-169.39 ± 22.38	-169.14 ± 16.87	-161.87 ± 27.99
Hip	104.33 ± 15.95	86.86 ± 38.01	95.87 ± 20.37	95.63 ± 10.50	93.76 ± 20.11

Table 21: OWS participant 3 angular velocity averages (degrees /s) and standard deviations at FS and FO for TL and perturbed gait events: ankle dorsiflexion (+) and plantar flexion (-), knee extension (+) and flexion (-), and hip flexion (+) and extension (-).

	Non-Braced			**Braced**	
	TL Left FS	**PERTFS**	**REC2FS**	**TL Right FS**	**REC1FS**
Ankle	35.52 ± 40.51	7.03 ± 30.94	49.04 ± 98.72	-51.15 ± 5.93	-6.79 ± 38.42
Knee	-66.03 ± 20.20	-7.60 ± 57.36	17.25 ± 160.06	16.70 ± 16.11	52.31 ± 31.59
Hip	-3.99 ± 7.41	-23.35 ± 26.79	-62.80 ± 80.12	-50.49 ± 11.18	-62.73 ± 21.31
	Non-Braced			**Braced**	
	TL Left FO	**PERTFO**	**REC2FO**	**TL Right FO**	**REC1FO**
Ankle	12.25 ± 50.63	-8.81 ± 77.02	-7.06 ± 84.74	-23.67 ± 12.32	-11.45 ± 30.62
Knee	-119.95 ± 14.14	-159.21 ± 31.60	-115.45 ± 37.95	-132.11 ± 21.46	-118.78 ± 32.63
Hip	87.18 ± 3.34	85.47 ± 13.04	81.27 ± 16.88	82.36 ± 14.71	57.84 ± 30.27

Appendix S: Knee Joint Angle Range of Motion for OWS Participants

Table 22: Knee joint angle ROM (degrees) in TL and perturbed stride conditions for OWS participants.

	Participant 1	Participant 2	Participant 3
Non-Braced Leg			
Left TL Stride	67.3	48.6	68.2
Perturbation Stride	68.1	33.4	62.0
Braced Leg			
Right TL Stride	67.3	51.7	36.0
Recovery 1 Stride	68.0	39.3	28.5

Table 23: Knee joint angle ROM differences (degrees) between TL and perturbed stride conditions for OWS participants.

	Participant 1	Participant 2	Participant 3
TL vs. Perturbation Stride	-0.8	15.2	6.2
TL vs. Recovery 1 Stride	-0.7	12.4	7.5

Appendix T: OWS Questionnaire Results

Question	Participant 1	Participant 2	Participant 3
Putting on the OWS	Neutral	Very Difficult	Difficult
Applying the straps	Neutral	Easy	Easy
Removing the OWS	Neutral	Difficult	Easy
Putting on your shoe	Very Difficult	Difficult	Difficult
Balance when using the OWS	Neutral	Very Easy	Neutral
Sitting when using the OWS	Neutral	Neutral	Easy
Stability during level walking	Neutral	Very Easy	Easy
Stability when walking on uneven ground	Neutral	Easy	Easy
Attitude toward wearing the OWS	Neutral	Neutral	Accepting
Is the OWS comfortable	About Half the Time	About Half the Time	All the Time
Walking ability when using the OWS SCKAFO, compared to walking with current orthosis	Somewhat Worse	Much Improved	Somewhat Improved
OWS appearance	Acceptable	Good	Acceptable
Effort when using OWS, compared to current orthosis	Somewhat More	Much Less	Somewhat Less
The weight of the OWS is	Heavy	Not Light or Heavy	Not Light or Heavy
OWS support	Same as Without Orthosis	Very Secure	Very Secure
Donning and doffing	Some Trouble	Cannot Do Unassisted	Some Trouble
Does the OWS make you perspire?	No	Yes	No
Does the OWS rub your skin?	No	Yes	Yes
What do you like best about the OWS?	That people are doing work in this area	Tremendous support on my right knee (braced)	Ankle support better than present brace

Question	Participant 1	Participant 2	Participant 3
What do you like least about the OWS?	Works off and on, cannot count on it at this point	Cuts into the back of my leg, very hot to wear. Piece that fits in shoe is too large and hard to walk on	Pump on side. Will be hard to put on long pants. If pump leaks, there will be a big mess as it did in test
Additional Comments	I think that there could be a better triggering device. Speed, I think is the wrong way to go... maybe a bubble device like a level	Good start	Great crew that I worked with ☺