Investigation of Contralateral Leg Response to Unilateral Stiffness Perturbations using a Novel Device

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Abstract— **The etymology of the word "Anthropos", the Greek word for Human, includes one of the defining characteristics of human beings, which is the ability to stand upright and walk. Locomotion is one of the human's most important functions that serve survival, progress and interaction. The force stimulus generated by the interaction of the foot with the walking surface is a vital part of human gait. Although there have been many studies trying to decipher the load feedback mechanisms of gait, there is a need for the development of a versatile system that can advance research and provide new functionality. Moreover, the role of the load feedback in inter-leg coordination during walking is still not well understood. In this paper, we present a series of studies that attempt to shed light on the role of load feedback on inter-leg coordination using a novel system, called Variable Stiffness Treadmill (VST). The device is capable of controlling load feedback stimulus by regulating the walking surface stiffness in real time. We first present the main functionality of the VST, focusing on the real-time closed-loop control of stiffness. Using perturbations of the treadmill stiffness on one leg of healthy subjects, we investigate the inter-leg coordination mechanisms, in body-weight-supported gait. Results show that ipsilateral stiffness perturbations, affect the contralateral (unperturbed) leg in body-weight-supported gait, while their effect is dependent on the timing of the induced stiffness perturbations. The developed system and experimental protocols are uniquely useful for gait research, can improve our understanding of gait, and create new avenues of research on gait analysis, walking robots and gait rehabilitation.**

I. Introduction

Locomotion is one of the human's most important functions that serve survival, progress and interaction. Gait requires kinematic and dynamic coordination of the limbs and muscles, multi-sensory fusion and robust control mechanisms. The force stimulus generated by the interaction of the foot with the walking surface is a vital part of human gait. While the effect of load feedback on gait has been an active field of study (for example [1]–[9]), there remains a question of which gait control mechanisms explain the change in gait kinematics and muscle activation.

Investigation of the role of afferent sensory feedback to gait control mechanisms usually involve sensory perturbations and the analysis of their effects. Various platforms and protocols have been used to investigate reflex mechanisms during different phases of the gait with the majority of the experimental protocols focusing on over-ground walking and dropping of the supportive surfaces at distinct gait phases [10]–[13]. Although most of the aforementioned studies focused on the effect of unilateral perturbations at the ipsilateral leg muscles, the bilateral response has also been studied [12]–[14]. During posture maintenance, experiments including powerful unilateral displacement of one leg produced bilateral responses both in adults [15], [16] and in healthy human infants [17]. Moreover, the load feedback as well as the length of specific muscles during walking has been associated with the muscular activations of the leg [1], [6], [10], [14]–[20]. However, all the previous studies failed to separate the mechanisms of gait from those of body weight support and balance. Most of the experimental protocols did not provide balance support. Therefore, mechanical perturbations and sudden load changes triggered mechanisms related to body balance and posture. The latter justifies the activation of inter-limb mechanisms and therefore explains bilateral leg responses. However, little is known whether this effect is exclusively caused by the mechanisms for body stabilization and balance maintenance, or if it is also brought about from inter-limb coordination mechanisms of gait pattern generators.

Nevertheless, there is a gap in our understanding of gait, which is related to how force feedback on the foot affects inter-limb muscle coordination when body balance is not disturbed. The identification of those control mechanisms will be valuable since it will elucidate neural pathways that mediate feedback signals and evoke muscular responses in both legs. This will help distinguish those mechanisms from others related to posture and balance and answer fundamental questions regarding gait. Therefore, the meticulous investigation of the role of afferent feedback in walking control mechanisms under certain and controlled conditions (treadmill walking, partial body-weight support, limited vestibular feedback, etc) can significantly improve our understanding of sensori-motor coordination in gait, and provide insight for the development of novel strategies for gait re-training.

In addition to the need for proper body support to reduce mechanisms for body stabilization and balance maintenance, a versatile system must be used that can provide a wide variety of sensory stimuli. In previous studies, researchers have utilized compliant surfaces as a means to investigate the effect of load feedback on gait [11], [21]–[24]. However, these designs do not allow for the compliance of the surface to be changed in situ. Moreover, there is no ability to exert a prescribed force perturbation to the foot in real time while a subject is actively walking on the surface.

In order to address the gaps left by other devices, a novel system, called Variable Stiffness Treadmill (VST), has been developed with several advantages over existing devices. First of all, the VST has a wide range of controllable stiffness

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– theoretically zero to infinite, but maintains high resolution. Second, it has the ability to actively vary and control the compliance of the treadmill surface within the gait cycle. Unlike previous devices, the VST is capable of creating any profile of stiffness during an experiment and throughout the gait cycle. Third, by measuring the displacement of the walking surface, we can not only estimate the load force exerted on the foot, but can also exert a force on the foot by adjusting the stiffness in real-time. The above elements allows the introduction of a plethora of force perturbations to the leg that are impossible to implement with current devices. The VST system constitutes the first mechanical device that can alter the walking surface stiffness in real-time, with high accuracy, resolution and robustness.

In this paper, we describe the VST system and give experimental validation for interlimb coordination mechanisms. The rest of the paper is organized as follows: Section II details the various components of the VST system along with a description of the experimental protocol. Section III presents the kinematic effect of unilateral stiffness perturbations on the contralateral leg with healthy subjects. Finally, section IV concludes the paper with a brief discussion and summary of the contribution.

II. Methods

A. VST System Components

The VST system allows us to investigate the effect of stiffness perturbations with greater versatility and functionality than other devices by combining a variety of components into one unique system. The device is shown in Fig. 1. The major components of the VST include a variable stiffness mechanism, linear track, force sensor mat, split belt treadmill, DC treadmill motor, counter-weight system, custom-built body weight support with two loadcells that measure the weight of the subject being supported by the system, rotary encoder, load cell for measuring forces exerted by the foot, and a feedback controller. Each component is important to the system for proper investigation of human gait mechanisms, and will be analyzed individually below.

1) Variable Stiffness Mechanism: The main novel feature of the VST is the ability to vary the vertical stiffness of the walking surface (i.e. treadmill), therefore controlling the force interaction between the walker and the walking surface. The capability of the VST to achieve a large range of controllable stiffness with high resolution comes from a novel variable stiffness mechanism. In its most simplified form, the variable stiffness mechanism is a spring-loaded lever mounted on a translational track, as shown in Fig. 2. The effective stiffness of the treadmill, located at a distance x from the pivot joint, is dependent on the coefficient of stiffness S of the linear spring and the moment arm r through which it exerts a force [25]. By design, S and r remain constant, therefore, the effective stiffness of the treadmill can be controlled by changing the distance x. The variable stiffness mechanism is shown in Fig. 1, part A.

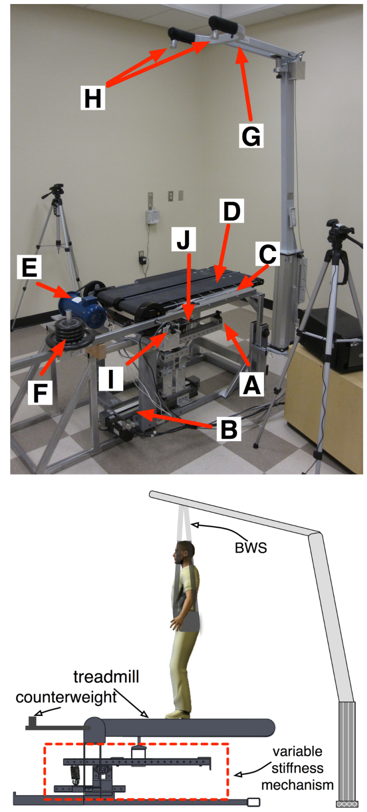


Fig. 1: The VST setup. Actual platform (top) and conceptual diagram (bottom). Subsystems shown include: A) Variable stiffness mechanism, B) Linear Track, C) Force sensor mat, D) Split treadmill, E) Treadmill motor, F) Counterweight system, G) Custom-made harness-based body-weight support, H) BWS Loadcells, I) Rotary encoder for treadmill inclination measurement, J) Loadcell for walker foot force measurement.

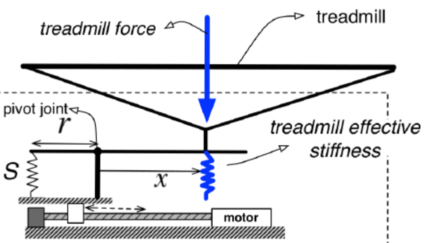


Fig. 2: Diagram of the variable stiffness mechanism.

2) Linear Track: The distance x is controlled by placing the VST mechanism assembly onto the carriage of a high-capacity linear track (Thomson Linear, Part Number: 2RE16-150537) which is controlled by a high-precision drive (Kollmorgen, Part Number: AKD-P00606-NAEC-0000). The resolution of achievable displacement of the linear track is $0.01\,mm$. Since the relationship between the linear track position and the treadmill effective stiffness in non-linear, the resolution of achievable treadmill stiffness is dependent on the linear track position. By performing a kinematic and kinetic analysis of the VST [26] we can compute the resolution for stiffness for any given linear track position. The resolution of stiffness can range from $9.06\,N/m$ when the linear track is at $5\,cm$, to $0.038\,N/m$ when the linear track is at its maximum displacement of $40\,cm$. Higher values for resolution are achieved for a position between 0 and $5\,cm$ of the linear track, as stiffness grows to infinity.

The range of the control of the track position defines the range of the treadmill effective stiffness that we can achieve. For $x_{track} = 0$, the treadmill stiffness is practically infinite, since the treadmill cannot be deflected. For the maximum displacement of the track of $40\,cm$, assuming that the foot of the subject is near the end of the treadmill (i.e. at toe-off phase), the minimum achievable stiffness is $61.7\,N/m$.

In addition to achieving the desired range and resolution of stiffness with the variable stiffness mechanism, we can also vary the treadmill stiffness actively throughout the gait cycle. In the most extreme scenario of going from a rigid surface, i.e. treadmill stiffness of $k_t = \infty$, to the minimum achievable stiffness, the linear track will have to move across its entire range (0 to $40\,cm$). Considering that the linear track can move as fast as $3\,m/s$, the system could make this extreme change in stiffness in $0.13\,s$. Assuming that the subject is walking at a normal pace of $1.4\,m/s$ [27], [28], with a stride length (the distance between consecutive points of initial contact by the same foot) of $1.4\,m$ [29], the stance phase would last approximately $0.5\,s$. This means that the variable stiffness mechanism can make this extreme change in stiffness three times during the stance phase. Therefore, it can easily change stiffness many times throughout the gait cycle when the desired change in stiffness is smaller than the two extremes. The high resolution for the adjustment of effective stiffness and the ability to change stiffness at a high rate throughout the stance phase of the gait cycle adds to the unique capabilities of the VST. The linear track is shown in Fig. 1, part B.

3) Force sensor mat: In order to track the location of the subject's foot, an array of eight force sensing resistors was placed beneath the treadmill belt. Whichever sensor is underneath the center of pressure of the foot will give the highest force reading. When two sensors give similar high force measurements, we can safely assume that the center of pressure is between the two sensors. So given that the sensor mat spans about $80\,cm$, with our eight sensors, we have a spatial resolution of $5\,cm$. Assuming that the average human foot length is about $23.5\,cm$ [30], this resolution is sufficient to know the location of the foot. The foot position is used as an input to calculate the corresponding linear track position that will create the proper apparent stiffness beneath the subject [26]. The force sensor mat is shown in Fig. 1, part C.

4) Split-belt treadmill: The VST employs a split-belt treadmill configuration in order to allow each belt to deflect different amounts. This will allow different force perturbations to be applied to each leg. The treadmill belts are supported at $70\,cm$ above the floor on a frame of steel tubing that permits each belt to independently deflect downward to a maximum of 30° from the horizontal position. The adjustability of the treadmill stiffness is currently limited to only one belt, but can be applied to both sides by installing another variable stiffness mechanism. The split belt treadmill is shown in Fig. 1, part D.

5) Treadmill motor: A 1-HP variable speed DC motor (Anaheim Automation, Part Number: BDA-56C-100-90V-1800) drives the treadmill belts. We can obtain speeds of up to $1.85\,m/s$ at a resolution of $7\,mm/s$. This includes the average preferred walking speed of $1.2 - 1.4\,m/s$ [27], [28], but can be slowed for individuals in therapy or rehabilitation applications. The treadmill motor is shown in Fig. 1, part E.

6) Counterweight: One necessary component to ensure accurate control of treadmill stiffness is a counterweight system to eliminate moments exerted by the treadmill's weight. This is achieved by fastening a weighted slider at the precise location along a co-linear beam which will induce an equal and opposite moment to that of the treadmill. This beam is attached to the side of the treadmill platform so that the counterweight system will cancel out the weight of the treadmill at any inclination of the treadmill. The counterweight is shown in Fig. 1, part F.

7) Body weight support: Separate from the treadmill structure, there is a custom-built body weight support designed by LiteGait. By adjusting the height of the support system, we can choose to have full or partial body-weight support. This is an important capability to reduce activation of body stabilization and balance maintenance mechanisms. In addition, the support increases safety and extends the system's capabilities to stroke patients and other individuals with decreased mobility and stability. Two loadcells attached on the body-weight support harness are measuring the subject's weight supported by the mechanism from each side. The body weight support and the loadcells are shown in Fig. 1, parts G and H respectively.

8) Rotary encoder: The angular deflection of the walking surface is measured with a rotary encoder (Encoder Products Company, Model Number: 260-N-T-11-S-1024-Q-HV-1-S-SF-1-N) in order to calculate the actual stiffness of the treadmill walking surface. The encoder has 1024 cycles per revolution resulting in an angular resolution of $0.35°$. The rotary encoder is shown in Fig. 1, part I.

9) Loadcell: The force exerted by the foot of the walker is calculated from the force measured by a 500 kg S Type Loadcell (RobotShop, Part Number RB-Phi-204) which is placed at the junction of the treadmill belt and the variable stiffness mechanism. This force is also used in the calculation of the measured stiffness of the treadmill. The loadcell is shown in Fig. 1, part J.

10) Feedback Controller: A hydrib controller, consisting of a Proportional-Integral (PI) feedback controller and a feedforward controller was designed and implemented in order to quickly achieve a zero steady state error of the actual treadmill stiffness in response to a desired stiffness reference signal. A block diagram representing the closed loop system is shown in Fig. 3 where k_t^d is the desired treadmill stiffness, err is the error signal, u_{FF} and u_{FB} are the feedforward and feedback control efforts, respectively, x_{track}^d is the desired

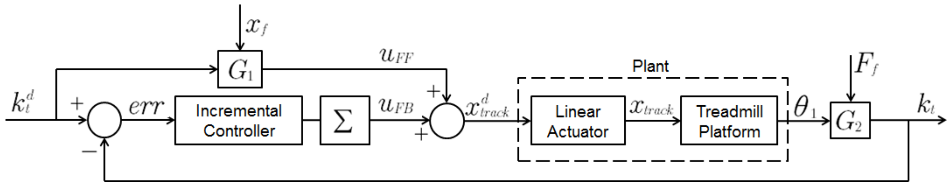
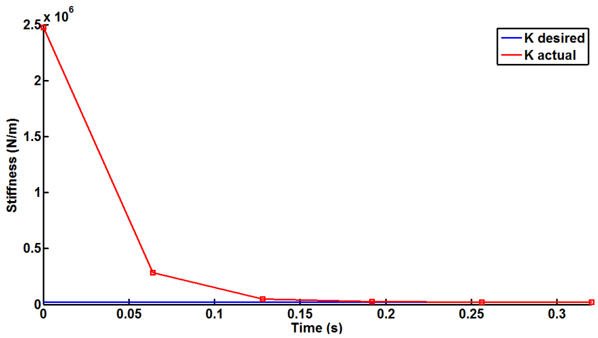
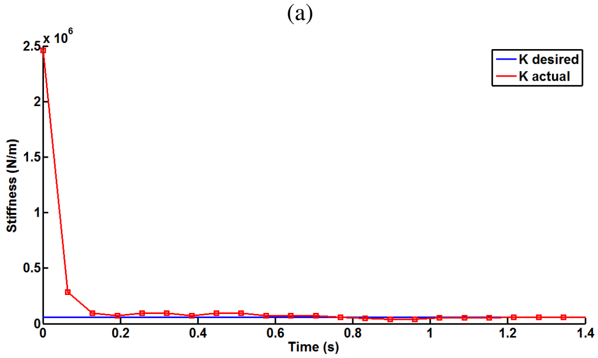


Fig. 3: Block diagram of closed loop system.



(a)



(b)

Fig. 4: Closed loop response to a step input of desired stiffness of a) 20 and b) 56 kN/m.

track position, x_{track} is the actual track position, θ_1 is the treadmill angular deflection, and x_f and F_f are the position and force of the foot, respectively. The transfer functions G_1 and G_2 are the results of a kinematic and kinetic analysis and describe the relationship of system parameters [26].

The feedback control structure was validated with two different reference stiffness values within the range of other variable stiffness devices [23], [24], [31]–[33]. A constant mass was placed at $0.33\,m$ from the treadmill pivot point and the desired stiffness was changed from rigid ($k_t^d > 2MN/m$) to 20 or $56\,kN/m$. The resulted stiffness is shown in Fig. 4. The rise time (t_r) and settling time (t_s) were calculated from a few repeated trials and are shown in Table I along with the steady state error. Due to minor fluctuations in the input variables (i.e. x_f and F_f) the steady state error has slight deviations from zero. However, this error is still only a fraction of one percent. We achieve a quick response and essentially zero steady state error validating that we can give accurate perturbations of desired stiffness for experimental investigation of gait control mechanisms.

TABLE I: Closed Loop Response

k_t^d (kN/m)	t_r (sec)	t_s (sec)	e_{ss} (%)
20	0.062	0.146	< 0.02
56	0.057	0.479	< 0.05

B. *Experimental Protocol*

We investigated the effect of a stiffness perturbation to one leg on the kinematics of the contralateral (unperturbed) leg while supplying appox. 30% body weight support in order to minimize balance and postural mechanisms without altering the gait kinematics by providing total support. Previous trials with healthy subjects demonstrate that 30

Six healthy subjects (3 female, 3 male) walked on the treadmill at a speed of $0.50\,m/s$. First, mechanical stops were placed under each treadmill belt to provide a rigid surface under both feet for 50 gait cycles to record the subjects normal gait patterns. Then, the mechanical stop was removed from the left treadmill and the stiffness was controlled to $1\,MN/m$, which is practically infinite, and does not differ from the stiffness with the mechanical stops. After a random number M of steps, where $M \in [3, 7]$, we immediately dropped the stiffness to $50\,kN/m$. The stiffness drop was done in 3 different stages of the left leg stance phase, (a) at loading response, (b) midstance, and (c) toe-off, as shown in Fig. 5. For the perturbed gait cycle, one out of the tree options were randomly selected. The low stiffness was commanded for about 40% of the stance phase (i.e. for approx. $450\,ms$), after which the stiffness was commanded back to $1\,MN/m$. An average of 15 perturbations per subject were done. The right leg was always supported by a rigid surface.

Kinematic data for both legs was obtained using a motion capture system (3D Investigator, Northern Digital) that was used to track 10 markers located at the torso, hip, knee, ankle, and toe (5 markers on each leg) in order to calculate the joint angles throughout the gait cycle. Cycles in between perturbations at infinite stiffness (except for two cycles following a perturbation) are included in the unperturbed data.

III. Results

The kinematic data of the contralateral leg in response to unilateral perturbations for a representative subject is shown in Fig. 6. The hip flexion-extension, knee flexion-extension and ankle dorsi/plantar flexion are shown (mean and standard deviation) across all gait cycles. The data is plotted as a

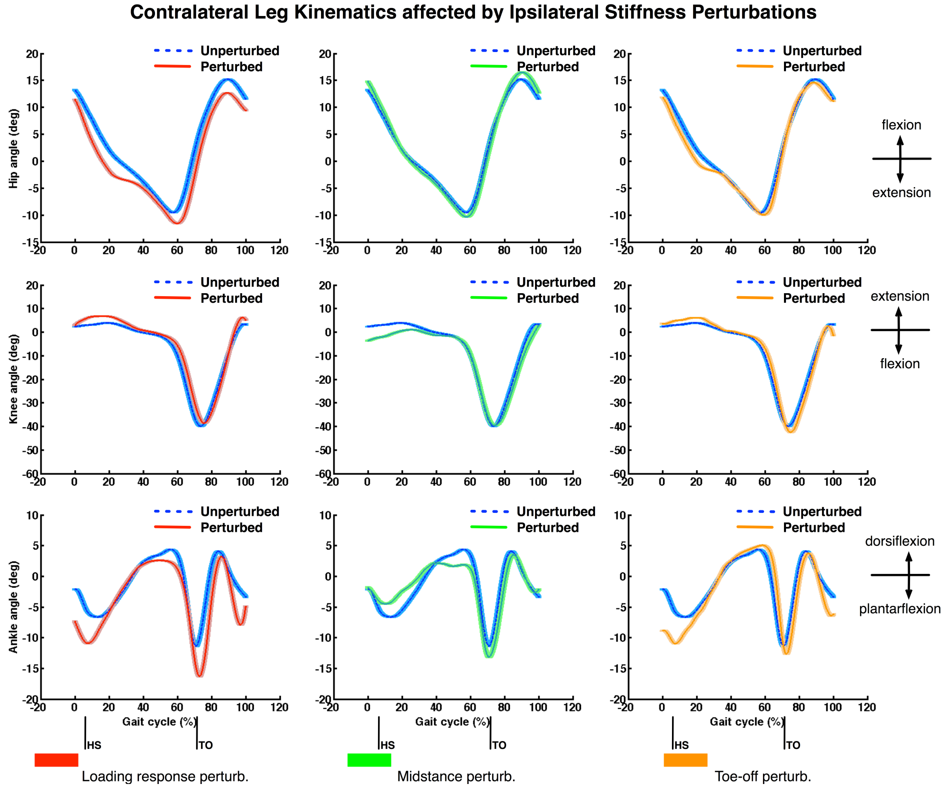


Fig. 6: Averaged kinematic data of right leg hip flexion (+) - extension (-), knee flexion (-) - extension (+) and ankle dorsi (+) - plantar (-) flexion for cycles with and without perturbation. Mean (darker lines) and standard deviations (lightly shaded areas) values are shown along with an indication of the timing of the perturbation. 100% of the gait cycle corresponds to approx. $2\,s$.

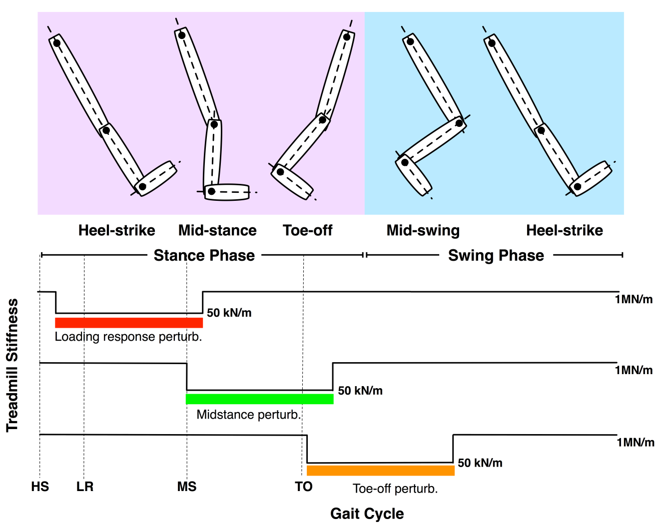


Fig. 5: Timing of unilateral perturbations

function of the gait cycle percentage, where heel-strike and toe-off are indicated on the figure as HS and TO, respectively. As can be seen, the joint angle profiles of the subject walking on a rigid surface resemble that of normal gait [29], [34], therefore our system did not alter the normal gait kinematics.

The plots also reveal an acceleration of right leg to contact the walking surface (right treadmill) in response to the stiffness perturbations on the left leg. This is shown in decreased hip flexion allowing for earlier contact with the walking surface and greater plantarflexion of the ankle joint due to the toe pointing down to make contact with the walking surface.

As shown in each subplot, there is a delayed response of joint kinematics after a perturbation compared to the unperturbed which is consistent throughout all joint angles and independent of the timing of the perturbation. In addition, the small standard deviation of the perturbed response profiles indicates that the effect of the perturbations was consistent and repeatable. The delay in joint kinematics ranges from a short (reflex-type) effect seen approx. $30\,ms$ after the left leg perturbation, to a late effect that can been seen at least $200\,ms$ after the perturbation. This observation results in the adoption of hybrid (spinal and supra-spinal) mechanisms involved in inter-leg coordination in body-weight supported gait. In other words, in addition to spinal Central Pattern Generators well discussed in the literature [35], the late response demonstrated by the kinematic data indicates the possible involvement of supra-spinal mechanisms of inter-leg coordination, that are centrally controlled.

IV. Conclusions

This paper presents the VST device that has been developed with several advantages over existing devices for gait research. The VST can alter the walking surface stiffness in real time, offering a wide range of available stiffness, practically from infinite stiffness (non-compliant walking surface) to as low as $61.7\,N/m$. Moreover, proper body weight support can be provided to reduce balance and postural control mechanisms allowing for a thorough investigation of sensory feedback on inter-limb coordination. Unlike previous devices, the VST provides a unique research platform to investigate gait control mechanisms.

This paper also presents results displaying the kinematic effects of unilateral stiffness perturbations on the contralateral leg, in body-weight supported gait. A systematic hybrid delay between the ipsilateral stiffness perturbations and the contralateral gait kinematics was characterized by short (reflex-type) effects seen approx. $30\,ms$ after the left leg perturbation, to late effects that can been seen at least $200\,ms$ after the perturbation. This observation results in the adoption of hybrid (spinal and supra-spinal) mechanisms involved in inter-leg coordination in body-weight supported gait. Muscle activation within the lower limbs will be investigated in the future to validate this result.

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