Anticipatory muscle responses in transitions from rigid to compliant surfaces: towards smart ankle-foot prostheses

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Abstract— **Locomotion is paramount in enabling human beings to effectively respond in space and time to meet different needs. There are 2 million Americans living with an amputation and the majority of those amputations are of the lower limbs. Although current powered prostheses can accommodate walking, and in some cases running, basic functions like hiking or walking on various non-rigid or dynamic terrains are requirements that have yet to be met. This paper focuses on the mechanisms involved during human locomotion, while transitioning from rigid to compliant surfaces such as from pavement to sand, grass or granular media. Utilizing a unique tool, the Variable Stiffness Treadmill (VST), as the platform for human locomotion, rigid to compliant surface transitions are simulated. The analysis of muscular activation during the transition from rigid to compliant surfaces reveals specific anticipatory muscle activation that precedes stepping on the compliant surface. These results are novel and important since the evoked activation changes can be used for altering the powered prosthesis control parameters to adapt to the new surface, and therefore result in significantly increased robustness for smart powered lower limb prostheses.**

I. INTRODUCTION

Locomotion is paramount in enabling human beings to effectively respond in space and time to meet different needs. In the US, an estimate of 2 million individuals live with limb loss and based on projections, this number is said to increase to about 3.6 million by the year 2050 [1]. Lower-limb amputations dominate, representing approximately 71% of dysvascular amputations [2].

Amputation of one or both lower limbs poses long term physical and psychological challenges for amputees with major issues relating to balance, falling and the fear of falling [3]. Standard prosthetic limbs are capable of restoring walking capabilities but are yet able to replicate natural walking in more complicated walking conditions.

Research indicates that approximately 52% of out-patients fall with major reasons relating to the prosthesis [4]. With the ankle joint being the most critical joint for gait stability and propulsion [5], [6], extensive work has been done regarding human gait to improve the design of powered ankle prostheses in dynamic walking conditions [7]–[10]. Adaptation to terrain is an important aspect of walking, and although previous studies utilize control methodologies for powered ankle prosthesis that have been very successful in walking and running [11]–[13], they generally fail to adapt to changes in compliance of the walking surface. Understanding how able-bodied humans integrate sensorimotor control mechanisms resulting in robust gait control in dynamic walking is essential for the design of advanced powered ankle-foot prostheses.

Limited joint angle mobility of the prosthesis of lower limb amputees, and the lack of distal muscles and sensory feedback from the lower limb results in difficulties while walking on uneven or non-rigid surfaces. Young, active transtibial amputees have been shown to increase toe clearance by increasing hip and knee flexion on the prosthetic side while increasing knee and ankle flexion on the intact limb during locomotion on a destabilizing rock surface [9]. Furthermore, Gates's study disclosed that variability of all step parameters and kinematic measures are affected by the surface type [8]. Intact individuals take conservative measures such as increasing minimum toe clearance to improve stability on complex surfaces and reduce the likelihood of falls. In addition, it has been demonstrated that there is a shift in the synchronization of muscle activation while walking on cross-sloped surface compared to level-ground walking. This reveals the existence of a possible feed-forward system for control as small alterations to walking surface were demonstrated to have significantly altered gait patterns [7]. Previous studies with human runners have shown that subjects adjust the stiffness of their stance leg to accommodate surface stiffness during steady state running, however did not indicate specific muscle anticipatory activation before the transition to the compliant surface [14].

While those previous studies are useful in understanding gait mechanisms involved in walking over some common obstacles, they were limited to hard, rigid surfaces, which only encompasses a limited type of natural environments individuals encounter daily. Additionally, studies carried out involving compliant surfaces identified the muscle activity only during walking on the compliant surface. All the previous studies have failed to identify kinematic and muscular activation pattern changes that result in transitioning from rigid to compliant surfaces. To advance state-of-the-art lower limb prosthetic devices, it is necessary that they can achieve performance levels seen in natural human walking. Despite the progress made in research findings, a gap remains in the ability of amputees using powered ankle prostheses for the maintenance of balance and stability when traversing

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complex, and especially compliant, terrains.

In this paper, we focus on the mechanisms involved during human locomotion, while transitioning from rigid to compliant surfaces such as from pavement to sand, grass or granular media. We hypothesize that significant muscle activity changes would be present just before and immediately after an individual encounters a compliant surface. Such changes can be used in the control of advanced powered ankle-foot prostheses to eventually achieve natural and robust walking on compliant surfaces for lower limb amputees. We utilize a unique tool, the Variable Stiffness Treadmill (VST), as the platform for human locomotion, simulating rigid to compliant surface transitions. The ability to simulate transitions between rigid and non-rigid surfaces, while measuring lower limb muscle responses, creates a window on sensorimotor control strategies for dynamic walking that has not been researched before. The results of our study provide solid evidence that when human subjects are prepared to transition from rigid to compliant surfaces, there exist significant muscle activity alterations that precedes the step onto the compliant surface. This result is very important since it can be used for altering the powered prosthesis control parameters to adapt to the new surface, and therefore result in significantly increased gait performance and stability.

The rest of the paper is organized as follows: Section II describes the experimental setup and protocol used for this study. Section III discusses the results of our study, while Section IV concludes the paper with a summary of the contribution.

II. Methods

A. Experimental Setup

The goal of this study is to analyze evoked activation of the lower limb muscles when human subjects expect to transition from rigid to compliant walking surfaces. For this reason, we conduct experiments where we apply expected unilateral stiffness perturbations and record activations from multiple muscles of the lower limbs. The walking surface stiffness perturbations are unilateral, i.e. at one leg, since in every-day walking, transitions from rigid to non-rigid surfaces, e.g. pavement to grass, are always experienced by one leg first, before the second leg transitions to the new surface. The investigation of muscle responses evoked by expected unilateral stiffness perturbations was performed using a unique tool, the Variable Stiffness Treadmill (VST) system shown in Fig. 1.

Briefly, the VST is a split-belt treadmill on which the compliance of the walking surface can be interactively and dynamically controlled. In its most simplified form, the VST is a spring-loaded lever mounted on a translation track that can change the effective stiffness under the foot by moving the linear track. An infrared camera system captures and monitors the location of the foot in real-time to control the timing of the stiffness perturbations throughout the gait cycle. The effective stiffness of each side/belt of the treadmill can range from its minimum value (61.7N/m) to its maximum, which is theoretically infinite (i.e., rigid walking surface), in 0.13s, which translates to 1/3 of the duration of a stance phase for walking at a normal pace of 1.4m/s [15], [16]. Moreover, the resolution of the VST stiffness control is approximately 0.038N/m [17]–[19]. These features allow for the introduction of a plethora of dynamic perturbations to the legs that are impossible to implement with current devices. The system has been detailed in previous work [17]–[19] and will not be described in this paper further for brevity.

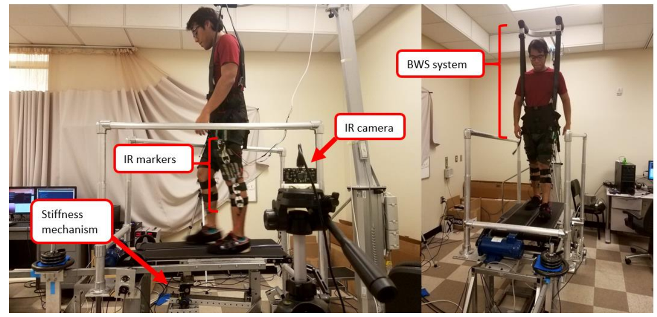

Fig. 1. The VST platform used in the experiments. The infrared camera system for tracking the leg motion is shown (IR camera, IR markers), along with the body-weight support (BWS) that was used for subjects' safety. The stiffness mechanism can alter the treadmill belt effective stiffness in real-time [17]–[19].

B. Experimental Protocol

Four healthy subjects [age 21.5 $\pm$ 1.9 years, weight 166 $\pm$ 37 lbs] walked on the VST at a speed of 0.60 m/s for at least 240 gait cycles. Since the goal of this experiment is to research muscle responses evoked by expected unilateral changes in surface stiffness, the right treadmill belt was not allowed to deflect (i.e. solid surface) for the duration of the experiment and subjects were informed verbally when the next change in surface stiffness would occur on the left treadmill belt. In other words, the changes in surface stiffness were effective for the left leg only and varied between rigid surface and compliant surface. For the rigid surface cycles the stiffness mechanism underneath the left leg was commanded to maintain a state for which it produces a very high stiffness (1MN/m) and considered to be rigid; for the compliant surface cycles the stiffness mechanism was commanded to produce a level of stiffness similar to that of a compliant surface, e.g. a gym mat, which is approx. 20KN/m. The timing of the change in surface stiffness occurred immediately after the left heel strike (approx. 5% gait cycle starting at the left heel-strike) and lasted for the duration of the left leg stance phase (i.e. until left leg toe off). The sequence of the changes in surface stiffness was: 30 sets of 1 rigid surface cycle, 30 sets of 1 compliant surface cycle followed by 3 rigid surface cycles, and finally 30 sets of 1 rigid surface cycle. Subjects wore a body harness for safety but no body weight support was provided. Informed consent from the subjects was obtained at the time of the experiment, and the experimental protocol is approved by the Arizona State University Institutional Review Board (IRB ID#: STUDY00001001).

C. Data Collection and Processing

1) Kinematics: Kinematic data for both legs were obtained at 140 Hz using an infrared camera system that is integrated with the VST [20]. The system tracked 12 (6 on each leg) infrared Light Emitting Diodes (LEDs) placed as pairs on the thigh, shank, and foot. The system provides the kinematics of both legs at the sagittal plane in real time. This data was also utilized for timing of the changes in surface stiffness.

2) Electromyography: The muscle activity of both legs was obtained using surface electromyography (EMG) via a wireless surface EMG system (Delsys, Trigno Wireless EMG) and recorded at 2000 Hz. Electrodes were placed on the tibialis anterior (TA), gastrocnemius (GA), soleus (SOL) and vastus lateralis (VL) of both legs. These muscles were selected as they play a primary role in ankle motion and stability, in which the TA produces dorsiflexion of the foot, the GA and SOL muscles produce plantar flexion of the foot, and the VL produces extension and stabilization of the knee [6]. After computing the EMG linear envelope, the data were normalized to the maximum value of each muscle. The EMG data corresponding to the gait cycles of walking on the rigid surface and the cycles pertaining to the compliant surface were found and categorized accordingly. Because muscle activity during walking is highly dependent on the phase of the gait cycle, the data were normalized temporally to percent gait cycle. The first 30 gait cycles and the cycles in between perturbations at rigid surface (except for one cycle following a stiffness change to eliminate any residual effects from the perturbation) are included in the rigid surface data set. This results in normalized EMG signals as a function of percent gait cycle, where 0% and 100% correspond to the heel strike of the left leg at two successive gait cycles. The statistical significance between the perturbed and unperturbed data sets was calculated at the 95% confidence level using an independent t-test.

III. Experimental Results

First, it is important to note that although the VST is commanded to alter the stiffness of the walking surface, it is not commanded to displace the treadmill belt vertically. The treadmill belt will be displaced vertically only when the subject steps on it, and the vertical deflection will be a function of the subject's load exerted on the belt, as well as its commanded stiffness. This is one of the unique features of the VST that makes it the only device that can mimic real life scenarios that involve walking on a compliant surface (e.g. grass). Therefore, when analyzing the responses of the human subjects to stepping on a compliant belt, henceforth called stiffness perturbation, it is worth analyzing it with respect to the vertical deflection of the treadmill belt, due to the loading of the belt and its lowered stiffness. The vertical deflection of the treadmill belt is measured via a high-resolution encoder mounted on the treadmill rotation shaft.

The treadmill vertical deflection for two gait cycles during the transition from rigid to compliant surface is shown in Fig. 2. The mean and standard deviation among 30 double-steps of unperturbed (solid surface) are shown in red dashed line. The mean and standard deviation among 30 double-steps of solid to compliant surface transition are shown in blue solid line. The first gait cycle in the perturbed case (blue solid line, 0-100%) corresponds to the step before the stiffness perturbation where the left leg steps on a rigid surface, while the second gait cycle (100-200%) corresponds to the step when the perturbation happens, i.e. when the left leg steps on a compliant surface. As is it seen, although the stiffness of the belt is commanded to change just before the beginning of the second cycle, i.e. just before the heel strike at 100%, the treadmill is significantly deflected after the heel strike of the left leg (at approx. 105%), when load is exerted on the compliant treadmill. Moreover, only downward motion is of importance, since the upward motion towards the end of the second gait cycle is due to the oscillatory behavior of the variable stiffness mechanism that can move upwards when the left leg leaves the belt and it is during the swing phase. This oscillatory behavior and upward motion is caused by the springs that produce the stiffness perturbations, as they are compressed during the stance phase of the left leg and released suddenly at toe off. This upward motion does not affect the experiment since the left leg is not in contact with the treadmill when this upward motion happens. Moreover the downward motion of less than 1cm that is shown before the perturbation (0-100%) is due to the elasticity of the device itself and although it is measured by the treadmill inclination encoder, it does not affect the experiment. Henceforth, we will consider that the stiffness perturbation starts at about 105%, when the downward deflection of the treadmill is significant and more than 1cm.

We choose not to analyze the leg kinematics in this paper since we expect that they will be very different between perturbed and unperturbed cycles, based on the vertical deflection of the perturbed belt. The kinematic changes generally do not precede the stepping on the compliant surface, therefore they can not be useful for anticipatory control for an ankle prosthesis. Moreover, in the case of the prosthesis, the natural ankle kinematics will not be available, therefore further analysis and use of the kinematic changes for the prosthesis control will not be pursued.

The EMG signals recorded from a representative subject are shown in Fig. 3. Figure 3 shows the normalized processed EMG signals from the four muscles of the left leg, the one that stepped on the compliant surface during the perturbed trials. The unperturbed (red dashed line) data show the mean and standard deviation double cycles of stepping on solid surface, for 30 double cycles. The perturbed (blue solid line) data correspond to the 30 double-steps of solid to compliant surface transitions. The first gait cycle in the perturbed case (blue solid line, 0-100%) corresponds to the step before the stiffness perturbation where the left leg steps on a rigid surface, while the second gait cycle (100-200%) corresponds to the step when the perturbation happens, i.e. when the left leg steps on a compliant surface. Based on Fig. 2,

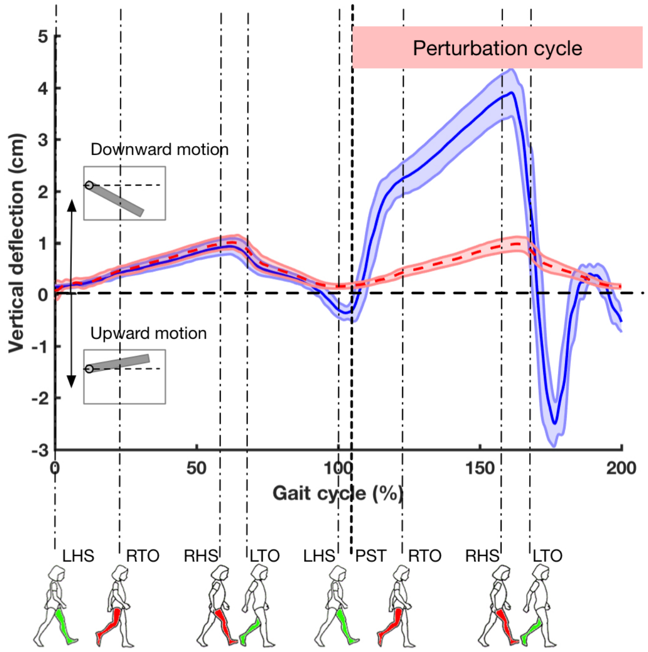

Fig. 2. Vertical deflection of the left treadmill belt across two gait cycles. Mean and standard deviation among 30 double-steps of unperturbed (solid walking surface) are shown in red dashed line. Mean and standard deviation among 30 double-steps of solid to compliant surface transition are shown in blue solid line. The first gait cycle in the perturbed case (blue solid line, 0-100%) corresponds to the step before the stiffness perturbation where the left leg steps on a rigid surface, while the second gait cycle (100-200%) corresponds to the step when the perturbation happens, i.e. when the left leg steps on a compliant surface. The transition from the rigid surface (first gait cycle 0-100%) to the compliant surface (100-200%) happens just after the left leg heel strike, at approx. 105%. This instance is named Perturbation Start (PST) on the graph. The gait phases of each leg are noted on the horizontal axis: Left Heel Strike (LHS), Right Toe Off (RTO), Right Heel Strike (RHS), Left Toe Off (LTO).

the perturbation starts at approximately 105%, as shown in the figure (PST vertical line). For each of the muscles VL, TA, GA and SOL, a statistical significance t-test is applied between the unperturbed (red dashed line) and perturbed (blue solid line) data for each of the data points throughout the two gait cycles plotted. If the difference between the data points is statistically significant at the 95% confidence level, then a magenta dot is plotted on the top of each sub-figure corresponding to each of the four muscles. As it can be seen, most of the muscles have statistically significant differences in the activation just after the perturbation (PST), which is more prominent at the TA, GA and SOL muscles. More importantly, there are differences in the muscle activation that precede the perturbation, i.e. that precede the step on the compliant surface. Those are more prominent at the VL, TA and SOL muscles, especially in the periods 70-90%, 75-105% and 75-95% respectively. These data provide strong indication that muscle activation encodes information about the preparation of the leg to transition from a rigid to a compliant surface. It is important to note here that the subjects were informed well in advance about when the compliant surface would be simulated, since they were in real-time provided with a countdown for the number of steps

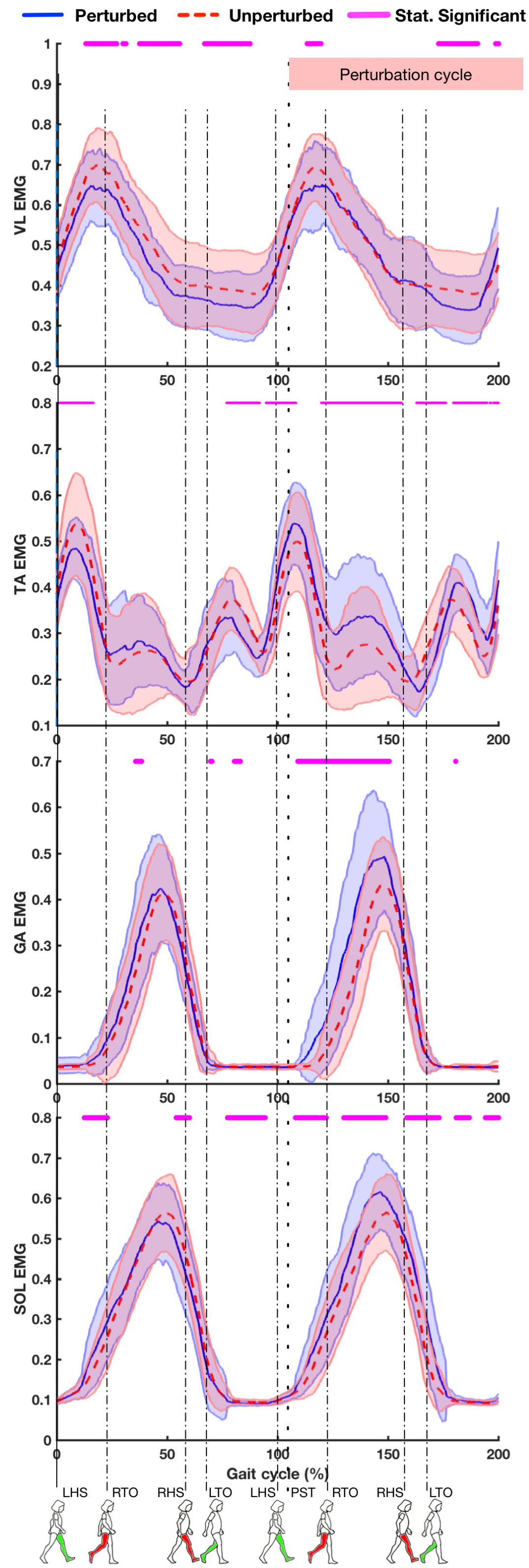

Fig. 3. Normalized processed EMG signals from the four muscles of the left leg. Statistically significant difference between the perturbed and unperturbed cases at the 95% confidence level is shown with a magenta dot plotted on the top of each sub-figure. The gait phases of each leg are noted on the horizontal axis: Left Heel Strike (LHS), Right Toe Off (RTO), Right Heel Strike (RHS), Left Toe Off (LTO).

before a compliant surface is experienced.

We also analyzed the contralateral muscles responses, i.e. the ones from the right leg to check if getting ready to step on a compliant surface with the left leg affected the right one as well. The EMG signals recorded from a representative subject are shown in Fig. 4. Figure 4 shows the normalized processed EMG signals from the four muscles of the right leg. The same information is plotted as in Fig. 3, which includes the result of the statistical test, as well as the timing of the perturbation on the left leg. Obviously the perturbation happens only on the left leg, and therefore when it happens (PST instance in the graph), the right leg is at the terminal stance phase, just before the right toe off (RTO). As it can be seen, the GA muscle of the right leg has statistically significant higher (and earlier) activation before the perturbation, at about 70-85% of the gait cycle. This provides indication that muscle activation on the contralateral leg can also encode information about the preparation of the leg to transition from a rigid to a compliant surface.

Finally, since four subjects participated in the experiments, it is worth analyzing the repeatability across subjects. We focused on the muscles of both legs that show statistically significant difference in activation before the perturbation. These are the VL, TA and SOL of the left (perturbed) leg, as well as the GA of the right leg. For those four muscles, the processed and normalized EMG signals mean for both the perturbed and unperturbed cases, across all trials are shown in Fig. 5 for each one of the four subjects. As it is shown, all four muscles of interest exhibit very consistent activation across the four subjects. Minor offset differences are expected due to the difference of muscle properties across subjects, but the overall patterns of activation for both perturbed and unperturbed cycles are seen repeatable.

IV. CONCLUSIONS

In this paper, we focus on the mechanisms involved during human locomotion, while transitioning from rigid to compliant surfaces such as from pavement to sand, grass or granular media. Our experimental study provides strong evidence that there exist anticipatory muscles responses on the lower limb when human subjects are prepared to step from a rigid surface to a compliant surface. More specifically, three muscles of the perturbed leg (VL, TA and SOL) show consistent differences in activation that precede the initial contact with the compliant surface. Based on our results, the TA and SOL show increased activation that comes earlier in preparation to step on the compliant surface compared to stepping on a rigid surface, while the activation of the VL decreases in preparation to the compliant surface. Those activation patterns could indicate increasing the ankle impedance before the transition to the new compliant surface, which has been noted in human runners in previous study [14]. Moreover, it was found that the GA of the contralateral (right) leg is also activated more and earlier when the ipsilateral (left) leg is prepared to step on the compliant surface. The results are repeatable across the subjects participated in the experiment. These results are very important since the evoked activation

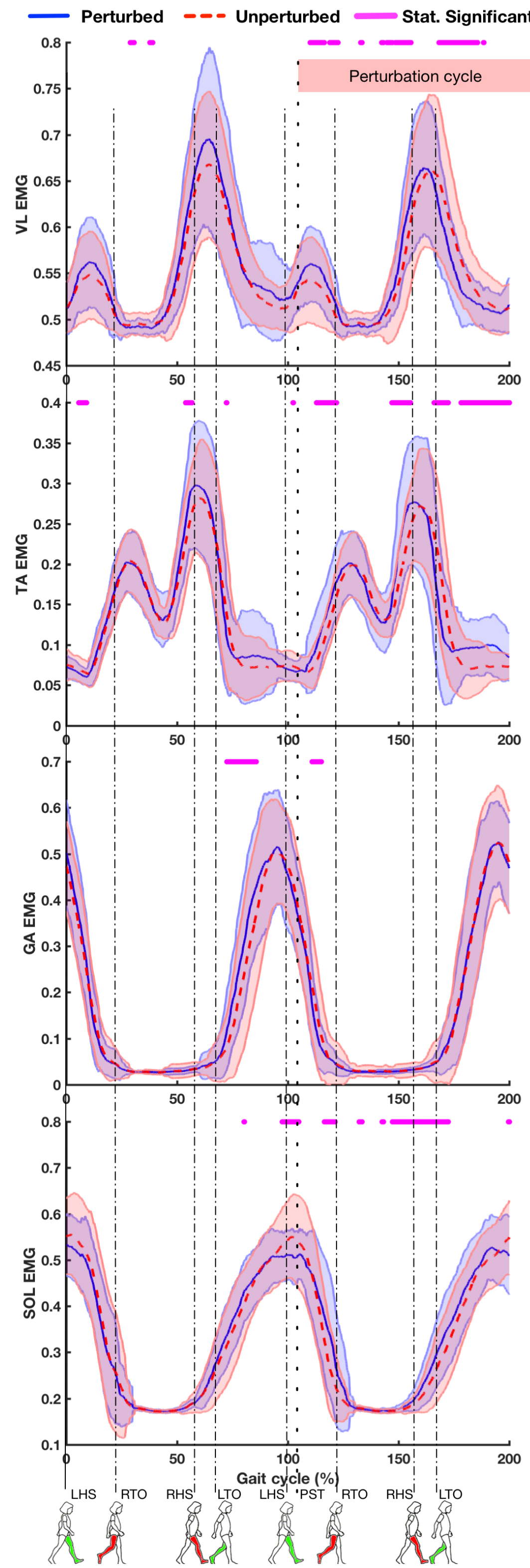

Fig. 4. Normalized processed EMG signals from the four muscles of the right (unperturbed) leg. Statistically significant difference between the perturbed and unperturbed cases at the 95% confidence level is shown with a magenta dot plotted on the top of each sub-figure. The gait phases of each leg are noted on the horizontal axis: Left Heel Strike (LHS), Right Toe Off (RTO), Right Heel Strike (RHS), Left Toe Off (LTO).

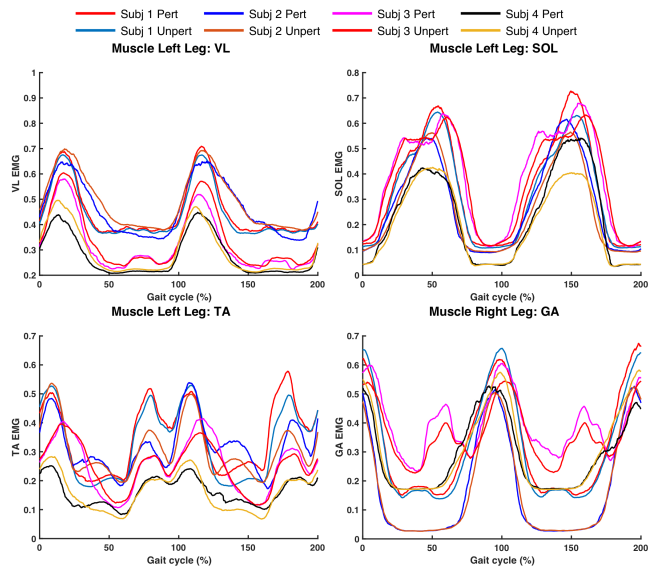

Fig. 5. Normalized processed EMG signals from three muscles of the left (perturbed) leg and one muscle of the right (unperturbed) leg, for both perturbed and unperturbed gait cycles, for all four subjects participated. Similar patterns are noted across subjects.

changes can be used for altering the powered prosthesis control parameters to adapt to the new surface, and therefore result in significantly increased robustness. The contribution of this paper can be found on the finding of solid evidence of specific muscle activity changes that precede stepping on a compliant surface, which can be used for developing smart controllers for advanced powered lower limb prostheses. Future work includes the utilization of those changes for the real-time adaptive control of powered lower limb prostheses.

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