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Quantifying Kinematic Adaptations of Gait During Walking on Terrains of Varying Surface Compliance

Lynsey D. Lehmann and Panagiotis Artemiadis*, *IEEE Senior Member*

***Abstract*— Locomotion is essential for a person's ability to function in society. When an individual has a condition that limits locomotion, such as a lower limb amputation, the performance of a prosthetic often determines the quality of life an individual regains. In recent years, powered prosthetic devices have shown nearly identical replication for human leg motion on non-compliant terrains. However, they still face numerous functional deficits such as increased metabolic cost and instability for walking on surfaces of varying compliance and complexity. This paper proposes joint angles of the biological leg are uniquely altered by surface compliance regardless of a subject's individual walking pattern. These differences are then displayed and quantified as a way to better characterize able-bodied walking compensation typical with three common terrains: sand, grass and gravel. This study also collects data outdoors using IMU sensors and is not limited by lab setup and conditions. These results are important since better understanding of joint angle kinematics on varying terrains could enable the formulation of advanced controllers for current prosthetic devices allowing them to anticipate surface changes and adapt accordingly.**

I. INTRODUCTION

Amputation due to disease and trauma affects the mobility and lifestyle of a growing number of individuals across the globe. In America alone, over two million people are living with a form of amputation with this number predicted to double by 2050 [1]. Nearly 90% of new limb losses concern lower extremities with 53% of patients requiring transtibial amputation [2]. Lower-limb amputation is a life-altering event, often resulting in patient depression, anxiety and failure to find enjoyment in previous activities [3]. As such, the performance of prosthesis often determines the range of opportunities and quality of life that an individual will regain [4].

Passive prosthetic ankle solutions have been around for decades, yet still fail to replicate the work and power production of the biological ankle [5], [6]. Studies with transtibial amputees have shown subjects use as much as 20-30% more metabolic energy to walk at the same speed as healthy individuals on non-compliant (hard) surfaces [7], [8]. Limited range of motion and rigid designs, as seen with the solid ankle cushion heel (SACH), restrict patient mobility and often result in various undesired gait compensations [9], [10].

Lynsey D. Lehmann is with the Mechanical Engineering program, at the School for Engineering of Matter, Transport and Energy, Arizona State University, Tempe, AZ 85287, USA llehman3@asu.edu

Panagiotis Artemiadis is with the Mechanical Engineering Department, at the University of Delaware, Newark, DE 19716, USA partem@udel.edu

*Corrresponding author: partem@udel.edu

Desire to more closely mimic normal gait patterns has encouraged the development of powered prosthetic ankle devices in recent years. These prostheses are programmable and equipped with numerous sensors capable of active position and impedance control [11]. A strong understanding of leg kinematics on hard, flat terrain has allowed powered prosthesis devices, such as Össur's Propio Foot[1] and Endolite's Elan[2], to very accurately mimic able-bodied walking and even running on surfaces like cement and asphalt. This success has been measured as an ability to reduce metabolic cost and match the joint angle and joint power profiles associated with an able-bodied biological limb [6], [12].

Even with these advancements, powered ankle prostheses still face major functional deficits for replicating walking on many non-rigid terrains. Previous work has attempted to record these shortcomings and examine amputee adaptations to current solutions. A study focusing on the effect of surface compliance on young amputees compared differences in walking trials for asphalt and long grass [13]. In comparison with intact subjects, transtibial amputees showed a significant difference in all metabolic and temporal gait characteristics for trials on tall grass. Walking speed was significantly slower and energy expenditure was greater for amputees, indicating that the subjects' prosthesis was not well adept for varying terrain. Another study having transtibial amputees walk across a destabilized rock surface further examined differences in walking adaptations between the intact and amputated leg [14]. Shorter, wider steps and increased toe clearance through asymmetric flexion of the hip and knee joints were observed. Asymmetry in joint angles for both stance and swing phase of the gait cycle reflected the amputee's challenge in maintaining walking speed while retaining overall stability.

These studies reflect challenges in creating adaptive powered prosthetic solutions that function equivalently to the lost limb. Failure of prostheses to duplicate the efficiency of the typical ankle or foot on any surface results in amputee concern with energy expenditure, asymmetric gait, instability, and extra stress on sound limbs [5], [15]. Better characterization of able-bodied human locomotion on various terrains could be useful in developing more adaptive and effective powered prosthetic solutions, thus minimizing gait compensation.

Part of the difficulty in characterizing the effects of terrain on able-bodied subjects is the challenge of performing exper-

[1] https://www.ossur.com/en-us/prosthetics/feet
[2] https://www.endolite.com/products/elan

iments outdoors on a greater variety of surfaces. Use of body worn inertial measurement units (IMU's) has shown recent success in measuring gait parameters in more locations, including outdoor settings [16]. A recent study using IMUs attached to the feet of able-bodied subjects had them walk on five unique outdoor terrains [17]. The results characterized the foot path and foot clearance of the subjects, allowing them to make conclusions on overall gait patterns for the terrains. Further investigations have similarly used IMU sensors on the foot to estimate foot trajectories on surfaces and demonstrate sensor accuracy in comparison to conventional motion capture systems [18], [19]. Yet, the foot is not the only indication of gait variation. Experiments having able-bodied subjects walk on a rock surface or uneven treadmills indoors show joint angle comparisons between terrains to be essential in determining trends in gait compensation [20], [21]. Thus, previous work indicates the need for outdoor experiments that closely analyze how leg joint kinematics are affected by walking surface properties.

This paper focuses on determining differences in joint-angle profiles of able-bodied subjects associated with three unique, compliant surfaces: sand, grass and gravel. We hypothesize that surface compliance directly affects gait kinematics and can be quantified by observing changes in the sagittal plane of the ankle, knee, and hip of able-bodied subjects at various key locations of a gait cycle. Information about able-bodied joint behavior on a variety of outdoor terrains could greatly improve controllers to accomplish powered prosthetic imitation of the biological leg, and be used as a tool to verify prosthetic performance. It could also be used as reference in other bipedal robotic applications.

II. Methods

A. Experimental Setup

The goal of this study was to examine differences in leg joint-angle kinematics for subjects traveling across a variety of compliant surfaces. Terrains to be tested were sand, long grass and gravel shown in Fig. 1. Outdoor locations were chosen both for the desired terrain and walking length of at least 18m. Distance carried importance to allow subjects to reach a steady gait and collect valid data uninfluenced from the start of the experiment. Conditions for long grass were blades of about 6cm in height. The sand surface chosen was that of a played-on volleyball court to closely resemble conditions in nature, such as at the beach. The gravel surface contained a variety of irregularly shaped rocks and stones sized from around 1-5cm. Subjects were instructed to walk on cement to establish a baseline for comparison to a non-compliant surface.

The YOST LABS 3 Space Wireless 2.4 GHz Inertial Measurement Units sensors (IMUs)[3] were used to conduct outdoor experiments and capture joint rotations. In addition to their portable, wireless functionality, IMU sensors have been proven to collect similar and accurate results to that of optical tracking [16]. The IMU sensors collected quaternion measurements during the experiments for eventual processing to joint angles.

[3]yostlabs.com/product/3-space-wireless-2-4ghz-dsss/

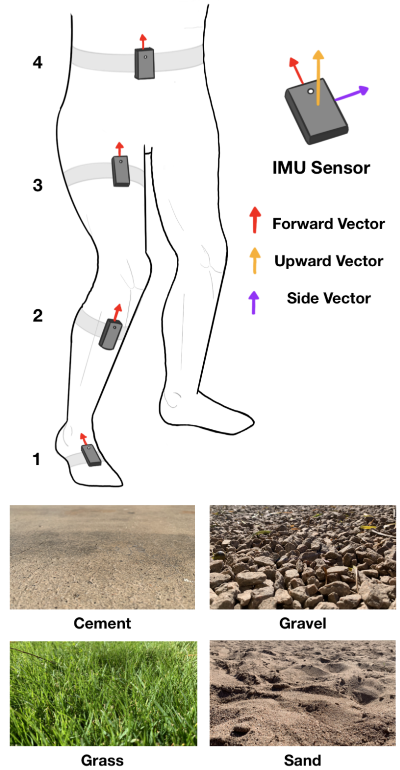

Fig. 1. Experimental protocol showing the location and orientation of IMU Sensors. IMU sensors were orientated with a left hand coordinate system with the forward vector of each directed up the leg and body. The four terrains tested are also shown from pictures taken at those locations.

B. Experimental Protocol

Nine healthy, able-bodied subjects participated in the study. All subjects gave written informed consent before starting experiments. Each subject was instructed to wear loose clothing, and a pair of athletic shoes to avoid restriction of movement. Four IMU sensors were placed tightly onto the foot, shank, thigh of a single leg, as well as the subject's torso by means of Velcro straps before the start of the experiment seen in Fig. 1. Sensors were carefully aligned on the frontal plane of the body before each trial.

Four terrains were chosen for able-bodied walking: cement (control), tall grass, gravel, and sand. Subjects were instructed to walk as normally as possible on the surfaces while following the pace of a chosen metronome at 100 beats per minute. This correlates roughly with subjects moving at 2.7 miles per hour. The beat was used for all subjects and helped them walk at a consistent velocity to avoid differences in joint angles based on speed.

Each subject walked for six trials on a single terrain. Trials ran for a total of one thousand data points collected for each sensor with a temporal resolution of 25 Hz. All IMU sensors were loaded off the subject and placed flat on the ground to reset their orientation and improve accuracy every two trials. Sensors were labeled and placed on the same body location each time.

C. Data Collection and Processing

Ankle, knee and hip angles were determined from measurements recorded by the IMU sensors with processing in MATLAB. Calculations regarding IMU sensor measurement to joint angles have been well documented by YOST Labs [22]. In short, quaternion measurements were converted into rotation matrices for all sensors, and the three (x, y and z) axes vectors were calculated with respect to their original tarred orientation. The z-axis ran parallel to the length of the subject's limb and was extracted from each sensor, referred to as the *forward vector*. Comparing forward vectors on the sagittal plane between two IMU sensors gave ankle, knee and hip angles of the leg. If $\mathbf{F_1}$, $\mathbf{F_2}$, $\mathbf{F_3}$ and $\mathbf{F_4}$ represent the *forward* vectors of sensors 1, 2, 3, and 4 respectively, then the angles of the ankle (θ_a), knee (θ_k) and hip (θ_h) on the sagittal plane are calculated by the following equations:

$$\begin{aligned}\theta_a &= Atan2\left(\sqrt{1-\|\mathbf{F_1}\times\mathbf{F_2}\|^2}, \|\mathbf{F_1}\times\mathbf{F_2}\|\right)\\ \theta_k &= Atan2\left(\sqrt{1-\|\mathbf{F_2}\times\mathbf{F_3}\|^2}, \|\mathbf{F_2}\times\mathbf{F_3}\|\right)\\ \theta_h &= Atan2\left(\sqrt{1-\|\mathbf{F_3}\times\mathbf{F_4}\|^2}, \|\mathbf{F_3}\times\mathbf{F_4}\|\right)\end{aligned} \tag{1}$$

The sign of the angles was determined based on constraints related to anatomical and range of motion for each joint, while zero-offsets were subtracted when the subjects were asked to stand still and straight before the experiment.

All joint angles per trial, terrain, and subject were manually observed and verified before further analysis. Each trial consisted of 15 to 25 gait cycles. Variation in number of gait cycles was due to length of the terrain, e.g. volleyball court versus grassy field. Trials with large amounts of noise due to various data collection and sensitivity issues faced by the IMU sensors were eliminated. All data was low-pass filtered for removing any noise, and processed data was compared to normal gait kinematics to remove any outliers [23].

With filtered trials, the representation of leg joint angles for all steps oscillating in time was converted into percentage complete of a single averaged step. Repeatability of local minimum between steps in the ankle was used to isolate gait cycles. Location of heel strike was determined as the point of local minimum following the largest amplitude in knee angle data for cement [23]. Finally, data was resampled using a cubic spline interpolation at 10 kHz to ensure all isolated gait cycles contained the same number of points for averaging and comparison purposes between terrains.

Following data processing, a t-test was used to confirm statistical significance between mean compliant vs non-compliant gait profiles at a 95% confidence for each of the resampled points. Differences in joint angles between surfaces were then extracted by a simple subtraction of all angles on the compliant surface from those on cement.

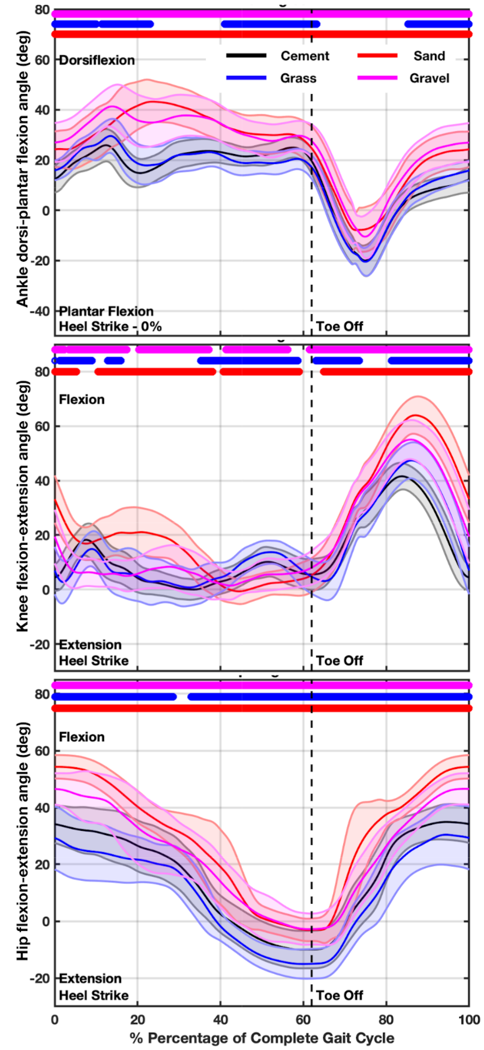

Fig. 2. Joint angle plots for the ankle, knee and hip of a single subject. Representative of the trends observed in the majority of subjects with clearly defined lines for each tested terrain shown to vary significantly from cement. Statistical significance (95% confidence) of variation from the control (cement) is noted by horizontal lines at the top of each subplot.

III. Experimental Results

A. Evaluation of Joint Angle Profiles on Compliant Surfaces

We found surface compliance to play a key role in defining the joint angle profiles of able-bodied subjects. Fig. 2 shows an example of one subject's profiles for the ankle dorsi-plantar flexion, knee flexion-extension and hip flexion-extension on all tested terrains. Mean joint angles across all trials are represented individually by the solid black, red, blue and magenta lines for cement, sand, grass and gravel respectively. The horizontal axis represents the percentage of a single gait cycle, with 0% representing the leg's heel strike.

Horizontal significance bars at the top of each joint angle plot identify areas of walking compensation. Colors correlate to the compliant surface used to perform the t-test with cement at 95% confidence. Length and position of the bars indicate trends in walking compensation at key locations of the gait cycle. It was common that walking compensations for a certain percentage of the gait cycle on one surface correlated to adaptations in that same range for other surfaces in a single subject. Comparing significance across subjects helped to identify the most likely areas of compensation in the gait cycle, and were saved in this phase for further analysis.

Assuming no two people can walk identically, variations in joint angle profiles between subjects were expected and examined. Comparing four subjects in Fig. 3, range and shape differences between joints likely correlated to factors such as walking posture, various eversions of the foot and control of the body's center of mass, among others. Observed differences included greater dorsiflexion in the ankle on sand and gravel during stance phase, and increases in knee flexion during the swing phase for all terrains among subjects 1-3. Select subjects, like 4, appeared to have greater control of their gait and walked almost identically to cement trials regardless of terrain. This was reflected in the significant decrease in significance bars above each averaged joint profile in comparison to subjects 1-3. Any areas of significance in these subjects were considered very useful information since it possibly highlights the most key sections of kinematic difference and walking variation.

B. Evaluation of Difference Plots

Difference plots were created to characterize the variation in joint angles for each compliant terrain. We hypothesized that the difference in joint angle kinematics between the compliant and non-compliant surfaces would objectively present trends in kinematic behavior of the joints. Hence, subject variation from factors such as slight differences in sensor placement and unique walking patterns did not affect comparison results.

Difference values were extracted between the compliant and non-compliant surfaces and plotted separately for all nine subjects as shown in Fig. 4. Plots are organized by the investigated joint and terrain, with black lines showing individual difference results per subject. Observed proximity of difference plot magnitudes and shape for all subjects reveal consistent trends in joint compensation. The colored lines in red, blue or magenta of each figure represent the averaged difference profile across all subjects. Values close to zero are indicative that gait kinematics are nearly identical to that of the cement surface.

Significance bars in Fig. 4 are representative of the number of subjects having the same locations of statistical significance as recorded from joint angle plots discussed previously. Hence, every point of the gait cycle correlates to a value 1-9 representing the sum of subjects with significance at the associated index. These results are plotted in green when at least six subjects have correlating indices, considered as the majority, and cyan for similarity between at least eight subjects. For the majority of subjects, critical locations of joint compensation encompassed nearly the entire gait cycle of all joints and tested surfaces.

To further compare joint angle trends between terrains, Fig. 5 combines all subjects and averages the joint angle kinematic differences. Discussions for Fig. 5 are referenced by the stance and swing phases of each gait cycle. Stance was considered in three phases: loading response (0-10%), mid stance (10-30%) and terminal stance to pre-swing (30-60%) [23]. Shaded error bars showing standard deviation are also plotted to indicate the possible variability in recorded results. Considering compliant surfaces to be ranked in order of least to most surface variation as grass, gravel and sand, thickness of standard deviation bars increases substantially on surfaces of greater variability. Furthermore, statistical significance correlating to at least six or more participants is represented by horizontal bars at the top of each plot.

Joint differences between terrains in Fig. 5 revealed important trends between surface compliance and ankle compensation both in the stance and swing phase of the gait cycle. Positive ankle angles close to heel strike indicated overall surface complexity, similar to shaded error bar thickness. Before each step, subjects showed to increase their dorsiflexion by around 5° to avoid surface contact during swing phase for sand and gravel. During midstance, increased dorsiflexion in the ankle appeared to represent the terrain's shape change under the force of the subject's leg. Values reflected observable deformation in the sagital plane as sand and gravel shifted around the location of the foot. Gravel also tended to hold the foot at an incline following heel strike and throughout mid stance. Large decreases in plantar flexion for sand, and a smaller decreases for gravel, are noted during terminal stance and pre-swing. This is significant since increased surface variability likely decreased subject confidence in placing weight in the foot before toe off. Ankle response on grass was unique in that the swing phase and terminal stance had slightly increased plantar flexion. Unlike sand and gravel, the grass surface acted like a spring absorbing and taking shape to the toe down motion of the foot before toe off. Otherwise, the remaining stance phase for grass was very close to cement.

Knee results gave further insight into walking behavior associated with each surface. Swing phase for all terrains showed significantly increased flexion around 95% of the gait cycle. Increased flexion of the knee was indicative of subject desire to increase toe clearance for each of the compliant surfaces [17]. Again, the magnitude of foot clearance correlated to surface irregularity and variation. Yet, surprisingly the magnitude of knee angle results were nearly identical for grass and gravel. Another interesting observation was the local minimum in the loading response of all terrains at around 8%. These values were likely a result of increased knee flexion at heel strike due to surface deformation, where full extension of the knee was finished during the stance phase. Greater magnitudes of knee angle flexion were also observed during mid-stance, possibly correlating with the perceived effort of the subjects to maintain stability in their gait on loose surfaces. Loading response and pre-swing showed decreases in knee flexion similar to trends in plantar flexion previously discussed for the ankle.

Hip observations were the final piece for understanding and observing the effects of surface compliance on able-bodied walking patterns. Positive differences in the swing

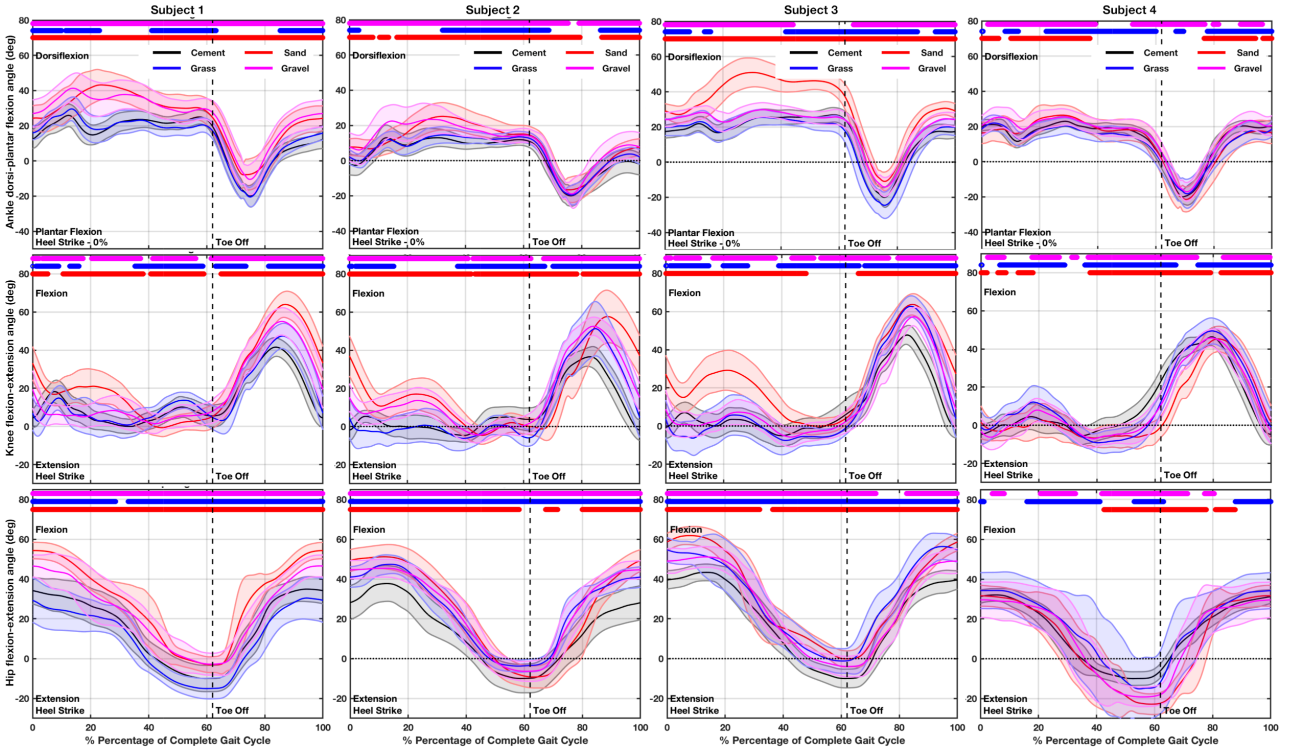

Fig. 3. Joint angle plots for four sample subjects. Trends between surface terrain and observed joint angles are easily observed in subjects 1-3 and more difficult to distinguish in subject 4. Statistical significance bars are shown on top.

phase indicated greater flexion and supported claims for increased toe clearance on compliant terrains. Extended range of flexion in hip angles carried over to loading response and mid-stance as well. Overall, larger hip flexion was likely reflective of greater energy expenditure needed to propel the body's center of mass back across the central axis of the hip. Hence, the difference values across tested surfaces reflected overall terrain variation and subject walking difficulty. Sand results at terminal stance were negative indicating reduced extension of the hip and less propulsion of the leg and body to the next step.

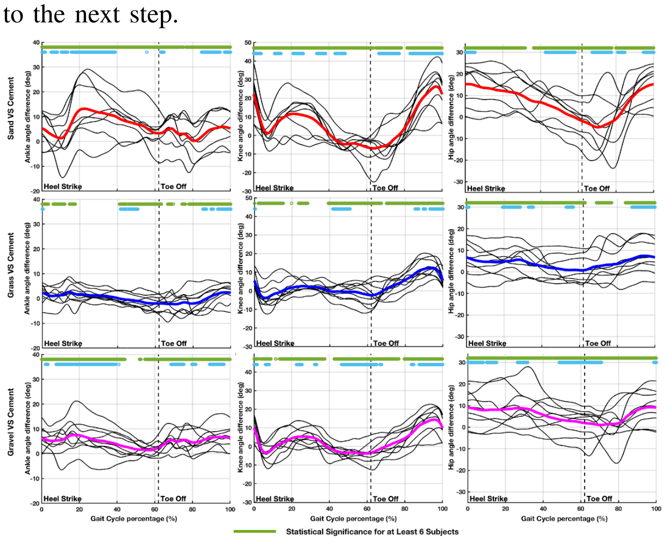

Fig. 4. Difference plots for all nine subjects shown by the black lines. Compliant surfaces and joint angles are separated to observe subject trends and the individual averages (colored lines). Statistical significance is plotted at the top of each plot for correlating the number of subjects the experienced adaptation at similar phases of the gait cycle.

IV. CONCLUSIONS

This study tested able-bodied subjects' walking on four unique terrains to determine differences in joint angle kinematics for the ankle, knee and hip. The experimental results provide strong evidence that surface compliance is critical in altering gait compensations made at multiple locations of the swing and stance phase of the gait cycle. Furthermore, difference plots quantify gait compensations, and show that more challenging terrains in regards to surface variability results in greater joint angle differences from a non-compliant comparison. Considering the order of surface variability from lowest to highest as grass, gravel, and then sand, the results are consistent in showing sand to have the greatest compensations.

Difference plots provide evidence for trends in walking kinematics on distinct terrains regardless of each subject's unique gait pattern. Statistical significance bars for the majority of subjects reveal that all the compliant terrains tested incur important changes in gait kinematics for nearly the entirety of the gait cycle. Assuming cement to be an example of walking under ideal conditions, quantified variation from that profile is extremely important to characterize able-bodied locomotion on a greater variety of surfaces. Understanding joint angle changes in able-bodied individuals is key in implementing improvements to powered prosthetic devices and other robotic applications to more accurately mimic bipedal walking.

The contribution of this paper is in determining gait compensations in relation to joint angle kinematics for walking on three very frequent, ubiquitous compliant surfaces. This research provides evidence that all compliant surfaces, even

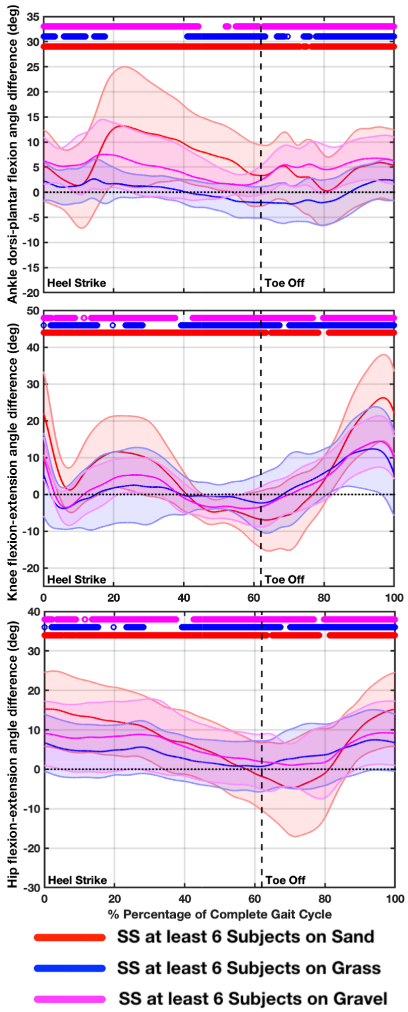

Fig. 5. Mean kinematic difference plots for all subjects with standard deviation. Statistical significance (SS) bars line the top of each plot for locations where at least six of the nine subjects had the same locations of SS. The bars represent locations of gait with important/expected joint compensation.

grass, introduce gait adaptations in the leg which are important when considering factors such as overall stability and walking efficiency. Future work could involve investigations for how joint angle adaptations vary with walking speed on a greater number of terrains, and implementation of results into real time adaptations made by a lower limb powered prosthesis.

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