Lower Limb Response to Modified Ankle Impedance in Gait

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ABSTRACT

This project used an exoskeletal robot to increase and decrease the stiffness of the ankle joint during treadmill walking to measure the effect of ankle impedance on lower limb joint kinematics. By quantifying the effect of ankle impedance on the knee joint we sought to better understand coordination and control of the ankle and knee. Using linear regression to determine the relationship between the maximum knee flexion during stance and the imposed stiffness on the ankle, we found a measurable positive correlation in 4 out of 5 test subjects at a 95% confidence level. The knee responded to modifications in ankle stiffness as expected from a simple mechanical model. Remarkably, the response was small and variable enough to suggest the body compensates to preserve normal kinematic profiles.

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1. INTRODUCTION

Gaining a quantitative understanding of how one piece of the human anatomy fits into a systematic picture provides insight into a holistic model of human physiology. We can use such understanding to better see how systems, such as the lower limbs in gait, can be manipulated and controlled. Information on the control of mechanical systems in the body, such as gait, can be used when designing therapies for persons with locomotor disabilities.

The work presented here seeks to understand the role the ankle plays in coordination of the lower limb joints in the human body during normal gait. This project robotically modified one mechanical aspect of the ankle during treadmill walking, and by monitoring the changes in the knee kinematics, we attempted to gain insight into knee control and gait adaptation.

Specifically, this project used an exoskeletal robot to increase and decrease the stiffness of the ankle joint and monitored the effect of stiffness on the motion of the ankle and knee. We found that the knee responded to modifications in ankle stiffness as expected from a simple mechanical model, but that the response was small enough to suggest the leg was adapting to the interventions to preserve normal kinematic profiles.
2. GAIT THEORY

2.1 Ankle and Knee Gait Kinematics

Gait is cyclic in nature and because of this, gait analysis often speaks in terms of a specific phase within a gait cycle. One gait cycle is broken up into the broad categories of stance and swing. Stance lasts from heel strike to toe off and swing lasts until the following heel strike. This breakdown is shown in Figure 1 where you can see stance lasting for about 60% of the gait cycle and swing accounting for the remaining 40% [1].

![Stance and Swing](image)

*Figure 1: One representative gait cycle. Stance is initiated by heel strike and swing is initiated by toe off. Reproduced from [1]*

Healthy knee and ankle angles during gait cycle have been well documented in the literature. Figures 2 and 3 below, taken from Perry’s text, *Gait Analysis* [1], describe healthy knee and ankle angle trajectories during walking. As in figure 1, the gait cycle onset is defined by heel strike.
Figure 2: Knee angle trajectory in healthy gait. Mean data is represented by the solid black line and +/- 1 standard deviation indicated by dashed lines. Zero degrees represents the fully extended leg. Reproduced from [1].

Figure 2 shows the normal motion of the knee during gait. Knee kinematics are characterized by 4 distinct motions: flexion post heel strike during limb loading, extension during early stance, rapid flexion once the contralateral leg has made ground contact (toe off occurs during this flexion—before peak knee angle is reached) and a final extension in preparation for heel strike [1]. Because the knee flexes following heel strike, we use this characteristic marker to estimate the timing of heel strike. Figure 3 shows normal ankle angular motion over a gait cycle.
The neutral standing angle of the ankle is at zero degrees in the above picture, dorsiflexion and plantar flexion correspond to positive and negative angles respectively. Two phases of interest are at heel strike when the ankle is near a neutral angle and then plantar flexes slightly as the foot rocks forward on the heel; and at approximately 60% of the gait cycle when toe-off occurs, at extreme plantar flexion. Here the ankle motion changes direction and begins to dorsiflex back to neutral as the foot leaves ground contact [1].

2.2 Mechanical Descriptions of the Ankle and Knee

Given a simplified model of the lower leg, as is represented in figure 4, predictions can be made about how the leg will respond to various interventions. Figure 4 represents the foot-ankle-shank system at the time of heel strike as 2 rigid beams connected by a simple spring and pivot joint. The system rotates about the heel, which is represented as a rigid extension of the forefoot.
Figure 4: Simplified model of the lower limb used to predict the effect of changing ankle stiffness: the foot-shank complex is two rigid beams connected by a pivot joint and spring. The limb pivots about the heel represented by a rigid extension of the foot at the time of heel strike.

While a model as simple as depicted above may not accurately model the ankle, it can be used to understand the effect of modifying the ankle joint stiffness. One could imagine as the spring becomes more stiff, it will exert a larger torque on the shank pulling the knee further into flexion, as the foot rotates about the heel pivot due to forward momentum and gravity forces transferred through the leg. Conversely, as the spring becomes less stiff, the knee of the system will not be pulled forward, rather the foot drops to the ground due to the transferred forces. This would result in hyperextension of the knee as the body mass maintains its forward momentum by pivoting about the ankle.

Passive ankle stiffness was measured robotically by Roy, et al. in 2007 [2], by applying an angular displacement and recording the torque required using MIT’s Anklebot (the exoskeletal
robot utilized by this project). Passive ankle stiffness was determined to be 9 N/rad in
dorsiflexion and 34 N/rad in plantar flexion, which agreed with previous literature [2]. However
the passive stiffness of the ankle does not fully describe the ankle during gait; the stiffness will
vary with the phase of gait cycle. Hansen, et al. [3] used motion analysis and a force platform to
measure the slope of the ankle angle versus torque curve taken from that data, thereby to
estimate ankle stiffness at different walking speeds. They found that at high and low walking
speeds, the ankle stiffness did vary during walking, but at normal walking speed the angle-toque
plots showed no net energy dissipation or gain, which is a behavior that could be mimicked (or
modeled) by an efficient spring [3].
3. METHODS

3.1 The Anklebot

This experiment made use of MIT’s Anklebot. The Anklebot is a robot designed for human rehabilitation that attaches to the test subject at the knee and foot [4]. The Anklebot is pictured below in figure 5.

![Figure 5: MIT's Anklebot. Reproduced from [4]](image)

The Anklebot consists of 2 linear actuators which, by means of ball joints connecting to the knee brace and Anklebot shoe, allow for 3 degrees of freedom of the foot relative to the shank. The knee brace was retrofitted with connections to the linear actuators, a shoulder strap for weight support, and a potentiometer for sensing knee angle. The Anklebot is designed to allow all possible movements of the ankle joint but controls only 2 of those degrees of freedom, dorsi-plantar flexion and inversion-eversion [4]. In this way it avoids exerting excessive forces due to poor alignment of the machine and the human joint.
The Anklebot is controlled in real time using torque and position sensors in the actuators and a linearized model of the Anklebot and ankle. The torque information is from an IMT current sensor with a nominal resolution of $2.59 \times 10^{-6}$ N-m. However, due to friction inherent in the system, the overall error in Anklebot torque was experimentally determined to be 1N-m. The resolution of the linear encoders that communicate with the controller is $5 \times 10^{-6}$ m. [4]

3.2 Safety and Human Factors

These experiments followed all National Institute of Health guidelines for the protection of human subjects in research. Informed consent was obtained from all test subjects before any experimental tests were performed. All test subjects were made aware of the most extreme forces the Anklebot would apply before being asked to begin the test and were aware they could ask for the test to be stopped at any point. These procedures were approved in advance by MIT’s Committee on the Use of Humans as Experimental Subjects.

As safety is a primary concern in all human studies, the Anklebot was originally designed with safety controls. The primary drive of the Anklebot actuators is a traction drive which can be overcome with a preset force. The Anklebot software also monitors all forces, velocities, and displacements and will shut down the Anklebot if any limits are breached [4].

3.3 Test Details

3.3.1 Use of Anklebot

This experiment used the Anklebot’s control system to define a rotational Anklebot stiffness and damping. We imposed a series of rotational Anklebot stiffnesses, both positive and negative, on healthy test subjects while they walked at their preferred walking speed. The imposed stiffnesses due to the Anklebot added to the overall stiffness of the joint.
The Anklebot control software defaulted to applying stiffness to both dorsi-plantar flexion and inversion-eversion. Adding negative stiffness in inversion-eversion resulted in the Anklebot attempting to roll the ankle. To maintain test subject safety, we modified the controller to apply the absolute value of stiffness to the inversion-eversion axis and only apply negative stiffness to the dorsi-plantar flexion axis. All analysis was confined to ankle dorsi-plantar flexion.

Primary data collection was from the Anklebot’s position sensors and the potentiometer on the knee brace. Anklebot software translates this data into ankle and knee angles. The Anklebot uses sensor data, ankle anatomy, and linearized foot-shank kinematics to estimate the ankle angle. The mean absolute error of the Anklebot’s ankle angle measurement was determined to be less than 1 degree in an experiment validating the use of the Anklebot as a clinical measurement tool [2].

3.3.2 Test Subject Preparation

Five subjects participated in this experiment: four males and one female. Subjects’ ages ranged from 22-26 years and none reported any history of neurologic or biomechanical disorders. The test subjects were outfitted with an Anklebot shoe and left leg knee brace as depicted in figure 5 in section 3.1. The knee brace was tightened as much as was comfortable and the Anklebot was attached while the subject was in a seated position. Upon standing the shoulder strap was attached and tightened to prevent knee brace slippage.

Test Subjects were then asked to walk on a treadmill while wearing the Anklebot (which was not actively controlled) to grow accustomed to walking with the Anklebot on one leg. During this acclimation phase, the preferred walking speed of the test subject was found. The test
subjects were asked to speed up the treadmill to the point where they felt comfortable walking for an extended period of time. Subjects were asked to increase treadmill speed until it felt too fast, then decrease it until it felt too slow, and repeat the procedure to ensure reliable estimation of preferred speed. This is a standard protocol used in the Newman Lab for Biomechanics and Human Rehabilitation and is described in Appendix A.

After determining preferred walking speed, test subjects were asked to stop the treadmill and stand in a neutral position. The linear encoders were zeroed at this neutral standing position. The stiffness of the Anklebot was then modified to the most extreme positive and negative stiffnesses. Subjects were asked to hold onto rails and move the foot attached to the Anklebot in order to feel the most extreme forces the Anklebot would apply to their foot while walking. Testing proceeded after ensuring the test subjects were comfortable with this range of Anklebot stiffness.

Before test subjects began walking, we recorded the knee angle at neutral standing (without locking the knee) and at a comfortable bent knee, for knee angle validation. The knee brace consistently measured 0 degrees when the brace itself was fully extended, but due to anatomical differences (and perhaps slight variations in knee brace placement), the knee angle of a standing test subject was NOT consistently recorded to be 0 degrees by the Anklebot system. There was an offset at neutral standing in knee angle that varied between 6 and 29 degrees across test subjects. We accounted for this offset by retroactively zeroing knee angle data at the neutral standing position. Because the knee at neutral standing is often slightly flexed, and this comfortable standing flexion varied across test subjects, some of our knee angle results show more hyperextension than is reported as typical in Perry’s text.
3.3.3 *Experimental Design*

This experiment sought to understand the effect of two basic types of ankle interventions: imposition of stiffness for an extended period, lasting over numerous strides, and a transient imposition of stiffness, applied only during an epoch centered on heel strike.

One extended intervention, lasting 25 seconds, was applied to the ankle at each modified stiffness magnitude. The extended intervention lasted between 15 and 22 strides, depending on test subject’s walking speed. This extended intervention was used to examine adaptation over multiple strides in addition to the ankle’s effect on knee angle.

Four transient interventions were applied to the ankle at each modified stiffness magnitude. These four transient interventions occurred 10-20 seconds apart. One transient intervention lasted 500 ms and encompassed 1 heel strike. The transient intervention was used to test our working hypothesis that the ankle stiffness primarily affects knee kinematics during the loading response following heel strike. To test this, we had to ensure the transient intervention always occurred during heel strike and the initial loading of the limb. We used the knee angle data collected from the Anklebot in real time to initiate the transient intervention shortly after a threshold knee angle (associated with swing) was crossed. One transient intervention and corresponding knee angle trajectory is illustrated below in figure 6.
Figure 6: Transient stiffness intervention about heel strike. Anklebot stiffness change is initiated after crossing a knee angle threshold and lasts 500 ms, thereby encompassing heel strike and the initial limb loading response.

The test consisted of approximately 15 minutes of treadmill walking while the stiffness was modified to 7 distinct magnitudes ranging from -15 to +45 N/rad. A summary of the stiffness schedule is presented in figure 7. The four transient interventions preceded each extended intervention.
Figure 7: Schedule of stiffness interventions. Neutral Anklebot stiffness was set to 2 N/rad. Anklebot stiffness was then modified with increasing magnitudes. Four transient interventions and one extended intervention occurred at each stiffness magnitude.

The most extreme stiffnesses reached were 45 and -15 N/rad. The maximum stiffness corresponded to an empirically determined maximum stiffness; this was where the compliance of the knee brace connection became a limiting factor. At -20 N/rad the Anklebot tended to become unstable for some ankle movements, so -15 N/rad was deemed a safe minimum stiffness. No test subjects experienced instability while walking on the treadmill.

As is shown in figure 7, the neutral stiffness for “unmodified” strides was set at 2 N/rad. This non-zero neutral ankle stiffness was used to compensate for the mass of the Anklebot system added to the lower limb inertia. Studies have shown that (as expected from basic mechanical dynamics) the resonant frequency of the human wrist joint is proportional to the ratio of joint stiffness divided by moment of inertia [5]. Calculations in ongoing work by Jooeun Ahn in the Newman Biomechanics Lab suggested that an Anklebot stiffness of 5 N/rad would
maintain a constant natural frequency of body sway about the ankle with or without the Anklebot. As we were concerned to study the effect of varying Anklebot stiffness, we used 2 N/rad to partially correct for the added inertia of the Anklebot while maintaining a reasonable range of variation above and below this value.

3.4 Data Analysis

3.4.1 Stride Division and Averaging

This analysis focused on two recorded values from the Anklebot’s data collection, knee angle and ankle angle in dorsi-plantar flexion during gait. Raw Anklebot data was sent through a FIR lowpass filter with a cut off frequency of 7.5 Hz to remove noise. Data was organized into six data sets: knee and ankle angle during the transient and extended interventions as well as unmodified knee and ankle angle data. Additionally, the modified strides were sorted based on the stiffness value. We identified the peak knee flexion for each stride and broke the knee and ankle data into individual vectors for each stride. Those vectors were then resampled to 100% of the mean unmodified stride duration in order to facilitate averaging. In MATLAB, resampling uses a filter to interpolate between data points. Because of the filter used, initial offsets needed to be removed to avoid “ringing” at the beginning and end of the angle data sets, and then replaced.

As mentioned in section 2.1, stride onset is typically defined by heel strike, where the phase of heel strike is 0% of the gait cycle, and toe off occurs at 60% of the gait cycle. Our working hypothesis suggests that modifying the ankle stiffness at heel strike will result in knee angle change. Because we modified the heel-strike event and may thereby have increased its variability, we chose to redefine 0% of gait cycle to be the peak knee flexion during swing which was less likely to be affected by our intervention. According to Perry’s normal gait cycle
definition, this onset would shift all events forward in the gait cycle by 30%. For instance, heel strike would move to approximately 30% of gait cycle and toe off to a little after 90% of gait cycle [1]. Using the peak knee angle during swing to define strides decreased the variability in duration of stride in one representative test subject from 3.75% of mean stride duration to 3.66% and more drastically decreased the variability in knee angle from 5.79 degrees to 0.84 degrees. A description of this stride definition is shown below in figure 8.

![Figure 8: Knee angle trajectory for 1 stride using peak knee angle used to define stride onset. With stride onset at peak knee angle, stance lasts from approximately 30% to 95% of stride.](image)

The phase of heel strike was estimated as the local minimum after the peak knee angle during swing and the phase of toe off was approximated as the minimum of ankle angle as discussed in section 2.1.
These strides were organized by sorting knee or ankle data into MxN matrices, where M was the number of strides in a given data set and N was 100% of gait cycle, in which \( n_{1-N} \) was the knee or ankle angle at the \( n^{th} \)% of gait cycle. These data matrices were used for averaging and determining the standard deviation of knee or ankle angle in a given data set at any given phase.

3.4.2 Peak Knee Flexion During Stance and Linear Regression

The aim of this research was to determine, first, whether ankle stiffness affected knee kinematics, and secondly, to quantify that effect. To establish whether there was an effect, we used the observation that the most obvious changes in knee angle occurred at the Peak Knee Flexion During Stance (PKFDS) and at the Knee Flexion During Heel Strike (KFDHS). Because the loading response in gait can be characterized by PKFDS, we focused the analysis here. We first found the phase at which PKFDS occurred for each stride. Occasional strides did not have a clearly defined peak during stance; in those cases, the knee angle at the average PKFDS phase was used. The average phase of PKFDS across all unmodified strides and all test subjects was 39% of gait cycle and the standard deviation of the mean PKFDS phase between test subjects was 3.3%. The phase of PKFDS for modified strides was never more than 1 standard deviation different from its phase in unmodified strides.

To determine whether PKFDS depended on ankle stiffness, all PKFDS values were linearly regressed against the associated stiffness intervention. Analysis of strides subject to extended and transient interventions for each test subject resulted in slope coefficients, 95% confidence intervals on that slope, and \( R^2 \) values to assess how much data variability the linear
If the 95% confidence interval on the slope coefficient did not encompass zero, the correlation between knee angle and ankle stiffness was deemed significant.

To determine which ankle stiffness interventions had a significant effect on knee angle, we performed a t-test to compare the PKFDS values for each stride subject to modified stiffness with the PKFDS values for unmodified strides. This tested the null hypothesis that the PKFDS values for strides subject to each of the stiffness interventions had the same mean as PKFDS values for unmodified strides. All test were conducted at a significance level of 5% or, equivalently, 95% confidence. For this test to be valid, the distributions of PKFDS values ought to be approximately Gaussian. To that end, histograms of PKFDS for representative data sets have been included along with the results of the test.

3.4.3 Adaptation of Strides Within Extended Intervention.

Finding adaptation in stride data could have interesting implications for rehabilitation. Because of this, preliminary knee angle data was visually inspected to assess any trend in strides subject to the extended intervention. The figures presented in section 4.4 showed encouraging results. In order to determine whether and how significant adaptation of knee angle data during the extended stiffness intervention occurred, we performed a linear regression vs. stride number for the PKFDS and KFDHS values for each stride subject to an extended intervention. This regression was performed for all test subjects at each stiffness. The regressions provided a slope of the trend of PKFDS and KFDHS values in time as well as the 95% confidence interval of that slope. Any data set that had a significantly positive or negative trend in the 95% confidence band for either of these knee flexion values was noted.
4. RESULTS

4.1 Anklebot data

Representative filtered Anklebot data is shown below in figures 9 and 10. Figure 9 shows 10 seconds of the knee angle trajectory in time, and has heel strike and peak knee flexion during swing indicated by markers. Figure 10 shows the same 10 seconds of ankle angle data.

![Filtered Knee Angle - Test Subject 4](image)

**Figure 9:** 10 seconds of filtered Anklebot knee angle data vs. time with heel strike and peak knee angle indicated by green and red markers respectively.
Figure 10: 10 seconds of filtered Anklebot ankle angle data vs. time with heel strike and stride onset (defined by peak knee angle) indicated by green and red markers respectively.

The red indicators defined by knee angle in figure 9 were used to parse the knee and ankle angle data into individual strides. All unmodified strides, and strides subject to extended or transient intervention were grouped. Figures 11 and 12 show the knee and ankle angle trajectories respectively for all strides in the unmodified data set. Figures 13 and 14 show knee and ankle angle trajectories respectively for strides subject to an extended stiffness intervention of $K=45$. Figures 15 and 16 show the knee and ankle angle trajectories respectively for all strides subject to a transient stiffness intervention of $K=45$. Mean values (solid black lines) along with standard deviations (dashed black lines) are plotted in each of these figures. These plots are consistent with Perry’s reported typical ankle and knee trajectories.
Figure 11: Knee angle trajectory of strides with unmodified stiffness. Individual strides are plotted in fine blue lines, the mean angle in the bold black line and +/- 1 standard deviation in the dashed lines.

Figure 12: Ankle angle trajectory of strides with unmodified stiffness. Individual strides are plotted in fine blue lines, the mean angle is plotted in the bold black line.
Figure 13: Strides subject to extended K=45 stiffness intervention. Individual strides are plotted in fine blue, the mean angle is indicated by the bold black line and +/- 1 standard deviation is represented by dashed lines.

Figure 14: Ankle angle of strides subject to extended K=45 stiffness intervention. Individual strides are plotted in fine blue, the mean angle in the bold black line and +/- 1 standard deviation in the dashed lines.
Figure 15: Strides subject to transient K=45 stiffness intervention. Individual strides are plotted in fine blue, the mean angle is indicated by the bold black line and +/- 1 standard deviation is represented by dashed lines.

Figure 16: Ankle angle of strides subject to transient K=45 stiffness intervention. Individual strides are plotted in fine blue, the mean angle in the bold black line and +/- 1 standard deviation in the dashed lines.
4.2 Stride Averaging

Figure 17 below shows the summary of plots like the ones above for knee angle data from the strides subject to all transient stiffness interventions. These data were from the same test subject as the above plots who was deemed the “best” test subject, or rather, the test subject with the most linear response to stiffness interventions. Each value of stiffness intervention is represented in a separate color. The average knee angle data for unmodified strides was plotted on top in black along with 1 standard deviation above and below in black dashed lines.

Following is figure 18, showing the average ankle angle data for the same strides. As you can see, the effect of the intervention appears to be small, but noticeable at the peak knee flexion during stance.
Figure 17: Colored lines represent average knee angle data of strides subject to transient stiffness interventions for 1 test subject. The black line is the average knee angle for unmodified strides which is surrounded by +/- 1 standard deviation between those unmodified strides.
Figure 18: Colored lines represent average ankle angle data of strides subject to transient stiffness interventions for 1 test subject. The black line is the average ankle angle for unmodified strides which is surrounded by +/- 1 standard deviation of ankle angle in those unmodified strides.

In contrast to the previous figures, the following figures, 19 and 20, show the data sets of knee and ankle angle for strides subject to extended stiffness interventions.
Figure 19: Colored lines represent average knee angle data of strides subject to extended stiffness interventions for 1 test subject. The black line is the average knee angle for unmodified strides which is surrounded by +/- 1 standard deviation between those unmodified strides.
Figure 20: Colored lines represent average ankle angle data of strides subject to extended stiffness interventions for 1 test subject. The black line is the average ankle angle for unmodified strides which is surrounded by +/- 1 standard deviation of ankle angle in those unmodified strides.

The ankle angle data for strides subject to extended interventions show a response to ankle stiffness interventions throughout the entirety of the stride, as would be expected. However the effect on knee angle is only seen during stance. During swing, when the ankle is no longer in contact with the ground, there is no visible effect on the knee angle.

4.3 Peak Knee Flexion During Stance - Linear Regression and T-test

As mentioned previously in section 3.4.2, two key phases in knee angle show the most effect: the peak knee flexion during stance, or PKFDS, and the knee flexion during heel strike, or KFDHS. As the loading response can be characterized by PKFDS the following analysis only
focuses on the knee angle at this phase. Figure 21 shows the linear regression of PKFDS for strides subject to extended intervention onto modified stiffness value. This data come from test subject 4 as above.

Figure 21: Linear regression of PKFDS for strides subject to extended intervention. Circle markers represent the PKFDS values for every stride subject to each stiffness value. The colors associated with stiffness values correspond to the same colors used in the stride averaging plots in section 4.2. The black line corresponds to the regression: $y=0.1217x+8.8807$. The 95% confidence interval on the slope is 0.0956 to 0.1478, significantly different from zero.
Figure 22: Linear regression of PKFDS for strides subject to transient intervention. Circle markers represent the PKFDS values for every stride subject to each stiffness value. The colors associated with stiffness values correspond to the same colors used in the stride averaging plots in section 4.2. The black line corresponds to the regression: $y=0.1978x+7.0996$. The 95% confidence interval on the slope is 0.1475 to 0.2481, significantly different from zero.

This type of linear regression was applied to each test subject. A summary of the slope coefficients (B), the 95% confidence interval of the slope, and $R^2$ values for the linear fit for each test subject is presented below in table 1. Results presented thus far have all been from test subject 4, who is noted in the summary tables.
<table>
<thead>
<tr>
<th>Subject</th>
<th>B</th>
<th>B 95% Confidence Int</th>
<th>R²</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>0.2331</td>
<td>0.1755</td>
<td>0.3428</td>
</tr>
<tr>
<td>2</td>
<td>0.0722</td>
<td>0.0296</td>
<td>0.0751</td>
</tr>
<tr>
<td>3</td>
<td>0.0398</td>
<td>0.0039</td>
<td>0.0384</td>
</tr>
<tr>
<td>4*</td>
<td>0.1217</td>
<td>0.0956</td>
<td>0.3690</td>
</tr>
<tr>
<td>5</td>
<td>0.2355</td>
<td>0.1934</td>
<td>0.5000</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>Subject</th>
<th>B</th>
<th>B 95% Confidence Int</th>
<th>R²</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>0.1215</td>
<td>0.0681</td>
<td>0.4572</td>
</tr>
<tr>
<td>2</td>
<td>0.0808</td>
<td>0.0148</td>
<td>0.1961</td>
</tr>
<tr>
<td>3</td>
<td>0.0306</td>
<td>-0.0713</td>
<td>0.0145</td>
</tr>
<tr>
<td>4*</td>
<td>0.1978</td>
<td>0.1475</td>
<td>0.7153</td>
</tr>
<tr>
<td>5</td>
<td>0.1300</td>
<td>0.0596</td>
<td>0.3569</td>
</tr>
</tbody>
</table>

Based on the above summary, we can say there is a measurable positive correlation between ankle stiffness and PKFDS because in all but one case the 95% confidence interval on the slope remains greater than 0. However, the R² values are all low, indicating high variability or a poor linear fit, so care must be taken when interpreting these results.

Two test subject’s linear regressions were particularly interesting for being atypical. Firstly, test subject 3 barely responded (or did not respond at all) to ankle stiffness interventions. As the test subject visibly walked more tentatively during the experiment, this could be explained by discomfort using the Anklebot. Secondly, test subject 1 responded more drastically to the extended K=-15 stiffness intervention than was expected. The following 2 figures show this drastic change in knee angle. The figures show a histogram of all PKFDS values, both modified and unmodified for this test subject. On top of these histograms a vertical line
representing the mean PKDFS for each stiffness intervention was plotted. The colors of the vertical lines correspond to the same stiffness as the stride averaging plots in section 4.2.

Figure 23: Histogram of PKFDS values for test subject 2, representative of the typical distribution for comparison. Mean PKFDS values for strides subject to extended intervention are shown in correspondingly colored vertical bars. The 2 bolded vertical black bars represent 3 standard deviations of PKFDS for all strides. Note that all modified mean values fall within the 6 standard deviation window indicating no group of data could be considered an outlier.
Peak Knee Flexion During Stance

Figure 24: Histogram of PKFDS values for test subject 1, an atypical distribution. Mean PKFDS values for strides subject to extended intervention are shown in correspondingly colored vertical bars. The 2 bolded vertical black bars represent 3 standard deviations of PKFDS. The pink vertical bar, indicating the K=-15 intervention, is more than 3 standard deviations away from the mean PKFDS value.

The discrepancy between PKFDS for strides subject to a sustained K=-15 intervention and the PKFDS values for all other data sets, indicated that the K=-15 data was somehow different from the remainder of the data set and should therefore be omitted from the regression analysis. However, since the only instance of such a pronounced effect was for test subject 1, the decision was made to keep all the data for all the analyses.

To determine what stiffness values could evoke a significant change in PKFDS we performed a statistical t-test to test whether 2 sets of PKFDS data might be considered as samples from the same distribution. A significant change was determined when the test negates
the null hypothesis that the 2 PKFDS data sets came from the same distribution, at 95% confidence. We performed this test for each stiffness, for strides subject to transient and extended interventions, compared with unmodified strides. With some exceptions we found that for the larger stiffnesses there was a significant difference between modified and unmodified PKFDS. A summary of how many test subjects showed statistically different kinematics between modified and unmodified strides is presented in table 2 below.

**Table 2: Number of subjects with statistical PKFDS differences at 95% confidence**

<table>
<thead>
<tr>
<th></th>
<th>K=-15</th>
<th>K=-10</th>
<th>K=3</th>
<th>K=5</th>
<th>K=15</th>
<th>K=30</th>
<th>K=45</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Extended Intervention</strong></td>
<td>4</td>
<td>3</td>
<td>1</td>
<td>3</td>
<td>2</td>
<td>3</td>
<td>4</td>
</tr>
<tr>
<td><strong>Transient Intervention</strong></td>
<td>4</td>
<td>1</td>
<td>0</td>
<td>2</td>
<td>0</td>
<td>1</td>
<td>1</td>
</tr>
</tbody>
</table>

This test assumes the data is normally distributed. To check the validity of using this test, figures 25-27 are histograms of PKFDS values. Figure 25 is a histogram of all PKFDS values for unmodified strides from test subject 2, a typical test subject. That data set was the comparison for all the t-tests for test subject 2. Figure 26 is a histogram of PKFDS values for strides subject to the extended K=45 stiffness intervention. Figure 26 is a histogram of PKFDS values for strides subject to the extended K=5 stiffness intervention.
Figure 25: Histogram of PKFDS values for unmodified strides from test subject 2. PKFDS values appear to be normally distributed.

Figure 26: Histogram of PKFDS values for strides subject to K=45 extended intervention. Data is from test subject 2. Given small sample size, PKFDS values appear to be normally distributed.
4.4 Stride Adaptation

We found suggestive evidence of adaptation of KFDHS and somewhat to a lesser degree of PKFDS and have shown this in figures 28-30. The data presented in the following 3 figures are individual strides subject to an extended stiffness intervention and were taken from the same test subject as was presented in sections 4.1 and 4.2. To visually inspect the strides subject to extended intervention each sequential modified stride was plotted in a slightly darker shade than the stride before it. The 3 figures represent different apparent adaptations. Figure 28 shows a
decrease in KFDHS, adapting towards the mean unmodified strides shown in yellow. Figure 29 shows a decrease in both PKFDS and KFDHS, but it trends away from the mean unmodified stride over time. Figure 30 shows an increase in KFDHS towards the mean unmodified knee angle.

Figure 28: Knee angle trajectory for strides subject to K=30 stiffness intervention. Sequential strides are progressively darkened with stride number to depict any trend over the duration of the extended intervention. The KFDHS decreases towards the mean unmodified stride which is represented in yellow.
Figure 29: Knee angle trajectory for strides subject to $K=-10$ stiffness intervention. Sequential strides are progressively darkened with stride number to depict any trend over the duration of the extended intervention. Both KFDHS and PKFDS decrease in time, but they trend away from the mean unmodified stride data at the end of the extended stiffness intervention. See figure 31 for a closer examination of KFDHS.
Figure 30: Knee angle trajectory for strides subject to K=-15 stiffness intervention. Sequential strides are progressively darkened with stride number to depict any trend over the duration of the extended intervention. PKFDS increases in time in these strides. Note that this trending is oppositely directed than figure 29, which are also strides subject to a negative stiffness intervention.

PKFDS and KFDHS were linearly regressed against stride number in order to more quantitatively measure adaptation. Linear fit slope constants and 95% slope constant confidence intervals were found. Figure 31 shows one example of this regression.
Figure 31: Linear regression KFDHS vs. stride number for strides subject to the extended $K= -10$ stiffness intervention for test subject 4. This regression corresponds to the strides seen in figure 29. The black line corresponds to the regression $y = -0.8511x + 4.2371$. The slope 95% confidence interval is -1.2428 to -0.4593, significantly different from zero.

Tables 3 and 4 provide the 95% confidence intervals for the slope coefficient for each set of data regressed to stride number. Confidence intervals greater than 0, indicating an increasing trend, were highlighted in green. Confidence intervals less than 0, indicating a decreasing trend, were highlighted in orange.
**Table 3:** 95% confidence intervals on slope coefficients for PKFDS against stride number

<table>
<thead>
<tr>
<th>Test Subject</th>
<th>K=45</th>
<th>K=30</th>
<th>K=15</th>
<th>K=5</th>
<th>K=-3</th>
<th>K=-10</th>
<th>K=-15</th>
</tr>
</thead>
<tbody>
<tr>
<td>1 Low</td>
<td>-0.2353</td>
<td>0.2861</td>
<td>-0.2511</td>
<td>-0.2225</td>
<td>-0.2896</td>
<td>-0.1670</td>
<td>-1.9432</td>
</tr>
<tr>
<td>1 High</td>
<td>0.0931</td>
<td>0.2857</td>
<td>0.2599</td>
<td>0.3484</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>2 Low</td>
<td>0.2655</td>
<td>0.4901</td>
<td>-0.1425</td>
<td>0.1258</td>
<td>-0.1073</td>
<td>-0.3283</td>
<td>-0.0851</td>
</tr>
<tr>
<td>2 High</td>
<td>0.0668</td>
<td>0.5864</td>
<td>0.3390</td>
<td>0.0582</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>3 Low</td>
<td>-0.0481</td>
<td>0.0504</td>
<td>-0.2150</td>
<td>0.0799</td>
<td>-0.4155</td>
<td>-0.1403</td>
<td>-0.2795</td>
</tr>
<tr>
<td>3 High</td>
<td>0.0791</td>
<td>0.4456</td>
<td>0.2749</td>
<td>0.4160</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>4 Low</td>
<td>-0.1558</td>
<td>0.3567</td>
<td>-0.2601</td>
<td>-0.1168</td>
<td>-0.0800</td>
<td>-0.4298</td>
<td>0.0632</td>
</tr>
<tr>
<td>4 High</td>
<td>0.1022</td>
<td>0.3535</td>
<td>0.3495</td>
<td>0.4160</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>5 Low</td>
<td>-0.3348</td>
<td>0.0656</td>
<td>-0.0712</td>
<td>-0.2304</td>
<td>-0.3754</td>
<td>-0.5925</td>
<td>-1.5674</td>
</tr>
<tr>
<td>5 High</td>
<td>0.2671</td>
<td>0.1673</td>
<td>0.3373</td>
<td>0.1321</td>
<td></td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

**Table 4:** 95% confidence intervals on slope coefficients for KFDHS against stride number

<table>
<thead>
<tr>
<th>Test Subject</th>
<th>K=45</th>
<th>K=30</th>
<th>K=15</th>
<th>K=5</th>
<th>K=-3</th>
<th>K=-10</th>
<th>K=-15</th>
</tr>
</thead>
<tbody>
<tr>
<td>1 Low</td>
<td>-0.1013</td>
<td>0.3507</td>
<td>-0.0275</td>
<td>-0.0980</td>
<td>-0.0984</td>
<td>-0.3754</td>
<td>-1.0289</td>
</tr>
<tr>
<td>1 High</td>
<td>0.1584</td>
<td>0.4461</td>
<td>0.5489</td>
<td>0.4614</td>
<td>0.4545</td>
<td>0.0994</td>
<td>0.0703</td>
</tr>
<tr>
<td>2 Low</td>
<td>0.2948</td>
<td>0.5811</td>
<td>-0.2131</td>
<td>0.2607</td>
<td>-0.0837</td>
<td>-0.3642</td>
<td>-0.1551</td>
</tr>
<tr>
<td>2 High</td>
<td>-0.2644</td>
<td>0.1735</td>
<td>0.1772</td>
<td>0.6723</td>
<td>0.4815</td>
<td>0.0446</td>
<td>0.2546</td>
</tr>
<tr>
<td>3 Low</td>
<td>-0.0290</td>
<td>0.4945</td>
<td>-0.2955</td>
<td>0.1124</td>
<td>-0.3097</td>
<td>-0.1599</td>
<td>-0.1501</td>
</tr>
<tr>
<td>3 High</td>
<td>-0.7501</td>
<td>0.4077</td>
<td>0.5026</td>
<td>0.3484</td>
<td>0.3484</td>
<td>0.5386</td>
<td>0.3861</td>
</tr>
<tr>
<td>4 Low</td>
<td>-0.4696</td>
<td>0.4067</td>
<td>-0.8349</td>
<td>-0.0683</td>
<td>-0.2151</td>
<td>-1.2428</td>
<td>-0.2001</td>
</tr>
<tr>
<td>4 High</td>
<td>-0.5755</td>
<td>0.0490</td>
<td>0.3476</td>
<td>0.5172</td>
<td>0.4593</td>
<td>0.7383</td>
<td>0.7383</td>
</tr>
<tr>
<td>5 Low</td>
<td>-0.6813</td>
<td>-0.1184</td>
<td>-0.0406</td>
<td>-0.0643</td>
<td>-0.4167</td>
<td>-0.7304</td>
<td>-1.1856</td>
</tr>
<tr>
<td>5 High</td>
<td>-0.1440</td>
<td>0.6674</td>
<td>0.5254</td>
<td>0.2650</td>
<td>0.3430</td>
<td>0.2641</td>
<td></td>
</tr>
</tbody>
</table>
Tables 3 and 4 do not reveal any obvious patterns regarding adaptation in either KFDHS or PKFDS. One would expect similar stiffnesses to have similar effects, even if only as a non-significant trend. As this was not the case, broad generalizations pertaining to adaptation cannot be supported by these data.
5. DISCUSSION

The goal of this work was to quantify the effect of ankle mechanical properties on lower limb joint dynamics. We used an exoskeletal robot to modify the stiffness of the ankle both transiently, only during heel strike and the loading phase of a gait cycle, and for an extended period of time, lasting for 17-20 strides depending on walking speed. Through linear regression, we quantified the effect of ankle stiffness on the peak knee angle during stance. This work has demonstrated that modifications of the ankle joint stiffness are measurably and positively correlated with peak knee angle during stance. We hypothesize that this may occur by the ankle stiffness pulling the knee further into flexion as the stiffness of the ankle is increased, inducing more rocking on the heel due to a lessened foot drop post heel strike.

The regression analysis on Peak Knee Flexion During Stance, indicated a measurable positive correlation for 4/5 test subjects but never with a R² fit value greater than 75% (and most frequently below 40% ). That is, the linear regression model could not account for more than 75% of the data variance (and frequently less than 40%). This indicates either that we cannot modify the stiffness of the ankle sufficiently to obtain substantial results, or that, while the physical mechanics may have some effect, the body compensates to preserve normal walking kinematics.

One of the test subjects did show a pronounced effect of stiffness intervention. Test subject 1 responded more than expected to the highest negative extended stiffness intervention. Given that none of the other tests subjects responded in the same manner, we are led to believe that this atypical response may be due to this test subject’s physiology. For instance, if the
highest negative stiffness is greater in magnitude than the relaxed stiffness of the test subject’s ankle, the net stiffness of the ankle joint becomes negative. As that would destabilize the ankle, it may evoke a different compensatory strategy such as antagonist muscle co-contraction to increase ankle stiffness and restore stability. However, without measurements of muscle activity, this is merely conjecture.

We tested to see whether the body adapts to preserve normal walking kinematics by calculating the change in 2 knee angles, PKFDS and KFDHS, over the duration of the extended stiffness intervention. The results of that analysis suggest that while systematic changes may occur over the course of 15-20 strides, there was not an observable consistent tendency to restore unmodified stride kinematics over that time course.

Future analysis might test for adaptation over a shorter time course using transient stiffness interventions, potentially varying in duration, and monitoring the trajectory of the knee angle throughout 1-2 strides.
REFERENCES


Appendix A:

Preferred walking speed protocol:

Find Preferred Treadmill Speed
Goal: pick a walking pace that is EASY & COMFORTABLE for period of about 15 minutes.
Procedure: You are going to walk at a comfortable pace, and then I will give you instructions to increase the pace until it is too fast, then too slow, and so forth to find a comfortable walking speed. You just have to follow my instructions and be comfortable

1. Start at a comfortable pace. After walking for [1 minute] increase the speed of the treadmill until you feel it is too fast to walk for a while
2. After [30 seconds] decrease back to a comfortable speed
3. After walking for [1 minute], decrease the speed until you feel it is too slow.
4. After [30 seconds] increase the speed back to a comfortable speed.
5. ASK: Is this comfortable?
6. RECORD speed
7. Rest [1 minute], then repeat