A Method for Quantifying Stiffness of Ankle-Foot Orthoses Through Motion Capture and Optimization Algorithm

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ABSTRACT Adjustments of Ankle-Foot Orthosis (AFO) stiffness is commonly prescribed for people with neurologically impaired to improve walking. It is important to quantify AFO stiffness levels to provide consistent patient-specific settings. Current AFO stiffness measurement methods require bulky, complex designs, and often permanent modification of the AFO. To address these issues, we propose the Ankle Assistive Device Stiffness (AADS) test method, a simple design jig using motion capture system and musculoskeletal analysis software (OpenSim). An AFO with three different but known dorsiflexion stiffness settings was installed to verify the AADS test method. Reflective markers were attached to the AFO while it was placed on force plates and four operators dorsiflexed the AFO with each stiffness condition, five times each. The collected marker trajectory data were imported to OpenSim to calculate AFO dorsiflexion angle using inverse kinematics. Then a static optimization algorithm was used to identify external forces from operators and the AFO torque best matching the experimentally collected ground reaction force data. Estimated AFO moments were compared within the operators’ trials and with theoretically calculated AFO moments to evaluate the accuracy of AADS tests. The AADS test results were repeatable through multiple trials and across operators. In low stiffness conditions, the AADS test had greater stiffness than actual results due to the friction of the AFO joint. As the spring strength increased, the AFO stiffness measured by AADS test was lower compared with actual stiffness due to the deformation of test AFO shells. The overall percent error between the theoretical and experimental stiffness was within ±6%. Moreover, the AADS test had high precision among the different operators and trials. AADS allows anyone with access to a gait lab quick and reliable AFO stiffness quantification. This is important to clinical practices, supporting patient-specific prescription and contributing to future research studies that require AFO stiffness.

INDEX TERMS Ankle foot orthoses, ankle assistive device, stiffness, optimization, OpenSim.

I. INTRODUCTION

Ankle-foot orthosis (AFO) is an assistive device intended to support neurologically or physically impaired individuals who require assistance in controlling and aligning their ankle and foot [1]. Adjusting dorsiflexion stiffness during stance phase is a common method for tuning AFOs that can provide subject-specific improvements in walking [2], [3]. The associate editor coordinating the review of this manuscript and approving it for publication was Jingang Jiang.

It has been shown that properly prescribed AFOs can reduce the energy cost of walking [4], [5], improve the gait pattern of the wearer [6]–[12] and provide stretching rehabilitation through gait [13]–[15]. Among many ways to tune the AFO to get the best outcomes, stiffness is one of the most commonly tuned parameters [16]. In current clinical practice, AFOs are prescribed largely by orthotist discretion rather than quantitatively which makes the process subjective [17]. Vasiliauskaite et al. found that using a quantitative method to customize the mechanical properties of AFOs to specific
To reach the optimal AFO stiffness we need reliable measurement methods. Typical AFO stiffness measurement methods use fixtures that affix the AFO to the testing device, a load cell that measures the engaged or response of torque, load, and potentiometers that measure the deformation of AFO [19]–[22]. There are many types of AFOs such as hinged joint AFO, posterior leaf spring AFO, and AFO with adjustable stiffness mechanisms. The variance of designs can create complex joint moments as AFOs are being dorsiflexed, capturing six degrees of freedom response (three translational forces and three rotational moments on each axis of the Cartesian coordinates system) that cannot be fully captured with uni-axial loadcells.

Using multi-axis loadcells with custom designed fixtures can provide accurate AFO stiffness measurements by capturing the moment from complex shapes and mechanisms [3], [20], [23], [24]. However, fixation of the apparatus affects the deformation of the AFO and loading responses. As a result, there might be discrepancies between the measured AFO stiffness designed with specific fixtures and AFO stiffness experienced in normal use [20]. The loading methods of previous measurement techniques are either manual or automated. Although automated loading methods provide convenient and consistent loading input, this requires controls of actuators leading to operation complex. As a result, these limitations hinders other research lab from adopting these methods [3], [20], [21], [25]–[27].

Using commercial torque or load measurement tools such as a human dynamometer [28], an industrial robot arm [29], or a universal testing machine [30] enables measuring the AFO stiffness without custom measurement device fabrication. However, these devices still require specifically design testing jig to apply bending to measure the AFO stiffness. In addition to this, the robot arm and universal testing machines are not directly related to the walking assessments with AFO, so it would be challenging to access these modalities for the people who analyze walking with motion capture systems. Both custom design AFO stiffness measurement apparatus and commercial measurement tools employ specific actuators, such as a motor, a hydraulic cylinder, and loading weights that require specially designed fixtures for the actuators, leading to heavy, bulky, and expensive testing devices.

To address current measurement limitations, we are presenting an Ankle Assistive Device Stiffness (AADS) test for determining ankle stiffness of AFOs through the use of a motion capture system rather than a specifically designed, complex, and fixated system. Using equipment readily found in gait labs, we derived an innovative technique that considers forces and moments on multiple axes to accurately measure AFO stiffness. In addition, using operator as a source of bending force instead of bulky and heavy actuators enables equipment to be compact and straightforward. Thus, this method allows accessible AFO stiffness measurement for both researchers and orthotists who are not trained to use specifically designed or commercially available measurement devices. To evaluate the feasibility and accuracy of the AADS test method, we tested AFOs with known stiffnesses to compare experimental results with theoretically calculated AFO stiffness.

II. METHODS

A. AADS TEST JIG DESIGN

The shank and foot used in the test was a 3D printed replica model of the subject’s leg. A hole was made in the shank to allow a 30 mm diameter pylon that is commonly used for lower limb prostheses to pass through vertically. The pylon creates a longer lever arm, reducing operator effort for higher stiffness AFOs. A ball and socket joint (CR-MO 1/2” DR., Lexivon, Moorpark, CA) was used to connect the pylon to the foot (Fig. 1). The AADS test jig design can be downloaded at https://simtk.org/projects/foot-orthoses.

B. ANKLE-FOOT ORTHOSIS (AFO) DESIGN

An individualized AFO, based on a 3D scan of a patient’s leg, was created by the 3D CAD Software, Meshmixer (Autodesk, Mill Valley, CA). As seen in Fig 1, the hollowed leg scan...
was trimmed to the shape of an articulated off-the-shelf AFO. A Camber Axis Hinge (Model 750, Becker Orthopedic, Troy, MI) connected the foot and shank portion on the medial and lateral sides. Extended strut bars were created on the posterior of the AFO shell to allow a spring attachment to provide dorsiflexion resistance. With the designed model, the fuse deposition modeling 3D printer (Pro2, Raise3D, CA, U.S.) was used to print the AFO. The thickness of the AFO was set to 3.5mm, and 3D printed using Advanced Polylactic acid (APLA) with a 70% infill.

C. TESTING

The trials for verifying AADS test performance, were conducted using twelve motion capture cameras (Vero, Vicon, UK) with a force plate (OptimaTM, AMTI, MA). From the motion capture cameras, marker trajectories on the AFO and AADS test jig were measured to identify the dorsiflexion angle of the AFO. The force plate was used to collect the ground reaction forces and moments in all three dimensions, as well as the center of pressure and angle measurements (Fig. 2). Reflective markers were placed following a modified Helen Hayes marker set [31]. Markers for the foot were placed on the AFO and leg replica in areas representative of the first and fifth metatarsal, middle cuneiform, and heel. Markers to represent the shank were placed at the equivalent location of the medial malleolus, the lateral malleolus, the anterior tibia, and the posterior tibia. The heel and posterior tibia markers were placed to track the displacement of the bottom and top of the spring respectively.

To start the trial, the AFO was placed on the force plate with the spring at the equilibrium position. Velcro was used to attach the foot to the force plate to prevent AFO heel lift and slipping. During the trial, the operators were instructed to push the AFO down, then return the AFO to the neutral position. The AFO has dorsiflexion stop limited to a maximum angle of 21° (Fig 3). Since the spring was not engaged at small AFO dorsiflexion angles, the torque of the AFO was obtained from the 5° dorsiflexion angle. To accurately determine the stiffness of the spring, torque out of range of minimum (less than 6°) and maximum AFO dorsiflexion angle (greater than 20°) were not considered for the stiffness calculations.

Three different springs were used to modify the AFO stiffness: 9,910 N/m (Spring 1), 15,600 N/m (Spring 2), 25,200 N/m (Spring 3). Each spring was randomly donned to the AFO and tested by four different operators, each running five trials. Each operator was given minimal instructions to ensure ease of use to future operators and validation of AFO stiffness repeatability in different bending conditions. From the instructions, the operator was able to decide the AFO dorsiflexion bending speed, direction of force, and range of AFO dorsiflexion angle.

D. MODELLING AADS TEST IN OPENSIM

OpenSim, an opensource musculoskeletal analyses software, was used to derive AFO stiffness [32]. A 3D model of the AFO and AADS test jig from the computer-aided design was imported to OpenSim to create a model. The model was composed with one rotational degree of freedom on the sagittal plane of the ankle joint, and six degrees of freedom on the center of mass of the entire AFO and AADS test models. The distance from experimental reflective marker positions and virtual markers in OpenSim were monitored to verify accuracy of scaling. The maximum marker errors for bony landmarks were less than 2 cm and overall root mean square error of all markers were less than 1 cm [33], [34]. The moment of inertia and the center of mass for each AADS test jig segment and AFO segment was calculated with SolidWorks (SolidWorks, Dassault Systmes SE, France). Also, each AADS test jig and AFO segment’s mass was measured with digital scale. These acquired moment of inertia, center of mass, and mass data were imported to the model in OpenSim.

E. DATA PROCESSING

Using OpenSim and the imported motion capture data from each trial, inverse kinematics was performed. Inverse kinematics calculated the AFO ankle joint angle by matching the experimental marker trajectories with the simulated marker set. To account for unknown external loads from the operator, static optimization (SOP) algorithm was employed [35]. SOP is designed to solve redundancy problems of human musculotendon dynamic systems. To customize SOP to AADS test analyses, three force and three torque actuators at the top of the shank model, and one coordinate actuator at the ankle joint were added. SOP determined the minimum input controls for these actuators that matched experimental reaction forces and moments from the force plate considering AFO kinematic. Using the estimated AFO torque profile with dorsiflexion angle, linear fit was used to determine the level of AFO stiffness in the sagittal plane.

F. VALIDATION OF RESULTS

The theoretical stiffness of the AFO was calculated to determine the validity of the results. During the trial, two markers were placed on the surface of the hooks (red circles in Fig 3). Using the position of these markers, a virtual marker was put at the proximal part of the spring \( T(x, y, z) \), and a marker was placed at the distal part of the spring \( B(x, y, z) \) (green circles Fig. 3). The spring length \( S_L \), was calculated by:

\[
|S_L| = \sqrt{(x_T - x_B)^2 + (y_T - y_B)^2 + (z_T - z_B)^2}
\] (1)

Having initial spring length \( S_{L0} \), the spring displacement \( \Delta S \) is derived by following equation.

\[
\Delta S = S_L - S_{L0}
\] (2)

Since we have used linear spring the force produced a spring is equal to:

\[
F = \Delta S \cdot k
\] (3)
FIGURE 2. An overview of the AADS test. Starting by applying users’ arbitral six degrees of freedom forces and moments to an AFO on a force plate, the AFO bends around its center of rotation. Using motion capture marker trajectory, inverse kinematics was processed in OpenSim. The results from inverse kinematics and experimentally obtained six degrees of freedom reaction forces and moments from the force plate were then used to find the AFO stiffness via static optimization algorithm.

The moment generated by spring, $M_s$, on the AFO ankle, was calculated by following equation.

$$M_s = F|\overrightarrow{AD}| \quad (4)$$

where $|\overrightarrow{AD}|$ is the normal distance between the ankle joint $A(x, y, z)$ and the line created by the $T$ and $B$ virtual markers. $\overrightarrow{AD}$ is connected to the line at point $D(x, y, z)$ at any given time. To find the position of point $D$, an unknown constant, $q$ were multiplied to the line created by the spring ($BT$), which $0 \leq q \leq 1$:

$$\overrightarrow{AD} = \overrightarrow{AB} + q\overrightarrow{BT} \quad (5)$$

$\overrightarrow{AD}$ is perpendicular to the $\overrightarrow{BT}$ dot product between two vectors crossing at a $90^\circ$ is equal to 0.

$$0 = \overrightarrow{BT} \cdot \overrightarrow{AD} \quad (6)$$

This was used to solve for the unknown variable, $q$:

$$0 = (x_T - x_B)(x_B - x_A + q(x_T - x_B)) + (y_T - y_B)(y_B - y_A + q(y_T - y_B)) + (z_T - z_B)(z_B - z_A + q(z_T - z_B)) \quad (7)$$

After $q$ was found, it was substituted into the equations solving for $(AD_x, AD_y, AD_z)$. The magnitude of this vector gives the moment arm, and was solved via:

$$|\overrightarrow{AD}| = \sqrt{AD_x^2 + AD_y^2 + AD_z^2} \quad (8)$$

Using equation (4) and (8), AFO stiffness is then calculated by:

$$k = \frac{M_s}{\theta} \quad (9)$$

where $\theta$ is the ankle angle from inverse kinematic. Although we have used linear spring, AFO stiffness showed some level of nonlinear behavior due to changes in moment arm. To introduce a representative constant number for AFO stiffness, we fitted a line to the torque-angle plot. This method was employed for every trial, and the average value for all trials on one spring was used as the theoretical stiffness.

G. STATISTICAL ANALYSIS

A multi-factor analysis of variance (ANOVA) [36] was used to determine which factor (e.g., different users and different springs) contributed the greatest to the total error to assess
the repeatability and reliability of the AADS test (Table 2). We defined different source of errors to verify which factor causes major error in accuracy of AFO stiffness estimation. Factors that likely caused the error are the different springs (Spring), the different operators (Operator), the different trials within the same operator (Trial), and the effect of the different springs on each operator that determines how varying spring stiffness could affect operators AFO bending performance (Spring × Operator). We also evaluated other factors outside the main experiment variable conditions (Residual). For example, an operator with less strength could struggle to bend an AFO with a stiffer spring. F-value (F) and Type I error (Prob > F) was calculated from the mean sum squared (MS), the partial sum squared (SS), and the degrees of freedom (DF) to evaluate how overall and each error source statistically impact the accuracy of estimated AFO stiffness with SOP.

### III. RESULT

Each AFO bending trial starts at $5^\circ$ dorsiflexion, the operators chose to bend the AFO angle with $15^\circ$ - $27^\circ$, depending on the operator’s discretion (Fig. 4). Both operator 1 and 2 bent AFO beyond the maximum dorsiflexion limit ($21^\circ$) in all spring conditions. Operator 4 only exceeded AFO dorsi-flexion in spring 1 condition. According to the SOP results, Spring 1 had a maximum torque of 16 Nm, Spring 2 had a maximum torque of 25 Nm, Spring 3 had a maximum torque of 36Nm.

In the Table1, the most compliant AFO stiffness case (Spring 1) had the smallest standard deviation (SD) out of the three springs. As the spring stiffness increased (Spring 2 and 3), the magnitude of the SD increased. The percent error of Spring 1 shows that the experimental stiffness value was 5.4% higher than the theoretical stiffness. As the stiffness of the spring increased, the percent error decreased in value. The experimental stiffness value of Spring 2 and 3 was 4.6% and 5.9% lower than the theoretical stiffness.

The F-value and the type I error values from multi-factor ANOVA demonstrates that the AADS test gives reliable and repetitive results regardless of different stiffness conditions, operator order, trial order, and each operators strength to control the AFO bending. Overall, there is no significant difference between estimated AFO stiffness from SOP and theoretically calculated AFO stiffness (Table 2).

### IV. DISCUSSION

The goal of the paper was to develop a novel method to determine the stiffness of an AFO. With the use of a force plate and motion capture cameras, the AADS test is quick, easy-to-use, compact, and an accessible method for those with a motion capture laboratory. The study showed the method is consistent across multiple trials and interchangeable between operators, suggesting the AADS test provides reliable AFO stiffness measures. While the theoretical AFO torque follows the same path on the plot when pushed down as when pulled up, there were hysteresis AFO torque curve from experimental results. The hysteresis curve was greater as spring stiffness and AFO dorsiflexion angle increases. This occurrence can be attributed to the AFO’s design and material property. Since the AFO is made of APLA, it deforms as a large force is applied to it [37]. At higher stiffness, more force is required to bend the AFO, causing more deformation. The viscoelasticity material properties of APLA dissipates the applied force from the operator, resulting in a lower stiffness. The decrease in torque can also be attributed to the AFO returning to its equilibrium position, as this would not need operator input. This AFO shell deformation contributed to an increased stiffness curve (Fig. 4).

The model in OpenSim is a rigid body segment and does not account for deformation of AFO shells. Ideally, the distance of the spring is proportional to the AFO dorsiflexion, however, the deformation of the AFO shank shell allows greater AFO dorsiflexion because the markers on the pylon and AFO shell move more in the interior direction while the markers on the AFO foot is constrained with force plates. The theoretical force is however calculated via the markers close to the joint center, meaning deformation will not influence the force value as much as the angle. As a result, the estimated stiffness from SOP would underestimate ground reaction forces and moment values. This is shown by the difference between the rising and falling experimental difference.

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### TABLE 1. A table summarizing the calculated stiffness for each spring in second column. Column 3 shows the mean and standard deviation (SD) of the static optimization results for 20 trials between the four users and the distribution of the trial results. In column 4, the deviation of mean stiffness of SOP from the calculated stiffness value is given.

| Spring | Theoretical Stiffness (Nm°) | SOP Stiffness (Nm°) | Percent Difference (%) |
|--------|----------------------------|---------------------|------------------------|
| 1      | 0.67                       | 0.71 ± 0.02         | 5.4                    |
| 2      | 1.07                       | 1.02 ± 0.03         | -4.6                   |
| 3      | 1.69                       | 1.59 ± 0.04         | -5.9                   |

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### TABLE 2. A summarization of the multi-factor ANOVA test.

| Source          | Partial SS | DF | MS  | F   | Prob > F |
|-----------------|------------|----|-----|-----|----------|
| Model           | 0.187      | 15 | 0.012 | 14.36 | 0        |
| Spring          | 0.175      | 2  | 0.087 | 100.31 | 0        |
| Operator        | 0.006      | 3  | 0.002 | 2.13  | 0.111    |
| Trial           | 0.001      | 4  | 0.000 | 0.35  | 0.845    |
| Spring × Operator| 0.006   | 6  | 0.001 | 1.17  | 0.34     |
| Residual        | 0.038      | 44 | 0.001 |       | 0.001    |
| Total           | 0.226      | 59 | 0.004 |       | 0.004    |
FIGURE 4. The graphs on the top show the plots from the static optimization results of different operators and multiple trials, while the charts on the bottom show the results from the theoretical calculations. For both the static optimization and theoretical graphs, all five trials of all four operators were plotted (20 trials per graph).

growing as the spring stiffness increases. This is also supported by the percent error becoming more negative as the stiffness grows (Table 1). Future studies should use a more rigid AFO to avoid these deformation issues.

While deformation of AFO shells led to the negative percent error in Spring 2 & 3, friction of the hinge joint was the major error source in the least stiff Spring 1. When operators applied force to bend the AFO, this force was applied to the hinge joint and created friction. The AFO shell does not deform with the low stiffness spring (Spring 1) as much as it does with the stiffer springs. Thus, the resistance from the friction of the hinge joint may profoundly overestimate the results. The AFO shell deformation and friction of the hinge joint contributed to the theoretical and experimental stiffness errors, causing a relatively larger MS value for different springs compared with the other sources (Table 2). When the same operator used the AADS test, the results showed consistent stiffness values. The MS value shows negligible variations in the stiffness across trials suggesting that the result of estimated AFO stiffness with SOP is independent of operator strength, as long as they can bend the AFO to the desired angle. These outcomes show that the method is repeatable via the same operator, among the same AFO, or different AFOs.

When including the entire trial, the error stemming from the different operators is higher than the other sources of error. This was due to the different maximum angles the operators decided to test. As operators bent the AFO, some users bent the AFO beyond 21° which was the maximum dorsiflexion angle for the AFO. As shown in the Fig. 2, the AFO stiffness of these cases had a greater stiffness slope compared with the AFO stiffness between 5° and 21°. This over bending of the AFO increased stiffness compared with the AFO stiffness attributed only to the spring. This led to greater discrepancies between operators, contributing to the F-value and type I error (Table 2).

As a previous study described, AFO stiffness varies depending on the alignment of the AFO rotational axis to the device [20]. This suggests that even if the same AFO is prescribed to each individual, the AFO applies a different stiffness due to inter-participant variability of rotational axes of the ankle joint on the sagittal plane and the frontal plane. Our AADS Test jig design is capable of adjusting the fore-aft and vertical position of the ball and socket joint on the Model-foot to match the wearer’s ankle rotation joint center. This versatile jig design will enable the measurement of wearer-specific AFO stiffness.

The study only tested the viability of the AADS test with linear stiffness AFO model. Unlike our AFO model, actual AFOs that have nonlinear stiffness profile such as posterior leaf spring AFO, double-action AFO, or other AFOs may have intrinsic nonlinear stiffness in their design. Our algorithm matches the ground reaction data and ankle joint angles at the hinge joint of AADS by modulating the user’s engaged forces and moments to find AFO torque using an optimization algorithm. This suggests that even if the AFO has a nonlinear stiffness profile, the optimization algorithm will find the solution of AFO torque and the user’s engaged forces.
regardless of nonlinearity. In addition to this, our induced nonlinear stiffness profile from the AFO shell deformation at a greater bending angle supports our method’s validity for AFO designs with nonlinear stiffness profiles.

The methodology also has the potential to measure the coronal and transverse stiffness of AFOs with multiple degrees of freedom on the joint because the AADS accounts for three dimensional forces, moments and kinematics. This suggests that AADS test can be used to measure the AFO stiffness that has complex posterior leaf spring, off-the-shelf trimline geometries. However, this study only validated sagittal plane stiffness of AFO due to the specific design of AFO. Thus, future study will validate the other planes of motion and nonlinear stiffness profile AFO. Due to the versatility of test method, the AADS test also has the potential to evaluate the comprehensive stiffness of prostheses, knee braces, and other gait assisting devices.

V. CONCLUSION
An easy-to-use, quantitative method of determining AFO stiffness directly impacts patient care. A quantitative measure of AFO stiffness will allow for tracking patient AFO prescription as the patient’s optimal AFO stiffness changes as their situation progresses. Having a procedure to measure the stiffness of the AFO can help orthotists prescribe and fabricate new AFOs for their patient. In addition, a record of the AFO stiffness will also be beneficial in refabrication when the AFO is damaged or requires resizing as the patient grows.

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