Design and Evaluation of a 2-DOF Instrumented Platform for Estimation of the Ankle Mechanical Impedance in the Sagittal and Frontal Planes

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Abstract—This paper describes the fabrication and initial evaluation of a vibrating platform with two degrees of freedom (DOF) to estimate the human ankle’s mechanical impedance in two DOFs; namely dorsiflexion–plantarflexion (DP) and inversion–eversion (IE). The device consists of an actuation and a force plate module. The actuation module generates torque perturbations up to 168 N·m in DP and 26 N·m in IE in the force plate module using Bowden cables. This provides a low-profile system that can be installed in a walkway. The frame of the force plate module rotates in two DOFs, applying torque perturbations to the human ankle in DP and IE. The ankle’s rotations are measured using a motion capture camera system. The analytical and numerical approaches for estimation of the ankle’s torques, rotations, and impedances are presented. A system validation using a mockup was conducted to verify the system’s ability to estimate the impedance of a physical system in two DOFs. The developed system was capable of identifying the mockup’s physical properties up to 15 Hz. The mockup’s impedance magnitude at 0.9 Hz using a stochastic identification method was shown to be within 1.68% and 0.54% of the mockup’s stiffness in DP and IE, respectively.

Index Terms—Ankle impedance, ankle kinematics, ankle kinetics, ankle moments, gait analysis, instrumented walkway, vibrating platform.

I. INTRODUCTION

The human ankle function is significant during walking and standing. During walking, the ankle supports the body weight, generates propulsion, and contributes in stability, while rotating and generating torques in dorsiflexion–plantarflexion (DP), inversion–eversion (IE), and internal–external–internal directions [1]. The ankle is also involved, among other functions, in balance and shock absorption. The mechanical impedance of the ankle changes continuously during a stride to accomplish this variety of tasks. The mechanical impedance of a physical system correlates an input motion to the system to the output reactive forces. In this paper, the mechanical impedance may be referred simply as impedance. The impedance represents the system’s resistance to motion as a function of the frequency of the input motion, or the relationship between effort and flow.

In a second order system, the impedance may be described as a function of the system’s inertia, damping, and stiffness. For the human ankle, the impedance is defined as the relationship between the angular motion disturbances to the ankle and its output reactive torques. The ankle impedance is both time-varying and task-dependent in the sense that it constantly changes during different phases of gait depending on the gait type [2], [3]. Although it has yet to be measured, to the best knowledge of the authors, there is strong evidence that the ankle impedance modulation is significantly different when comparing straight walking to turning maneuvers such as step turn or sidestep cutting, and also when comparing different gait speeds [1], [4]. This paper presents the design and the results of the initial evaluation of a novel platform that potentially allows for estimation and comparison of the mechanical impedance of the ankle in the sagittal and frontal planes (in DP and IE, respectively).

The human ankle’s capability to generate net positive work during gait and the ankle’s impedance at low frequencies (around 1 Hz, which is closely equivalent to the ankle’s stiffness) have been the core concepts in the design of powered ankle-foot prostheses. Such an approach allows amputees to benefit from prostheses that have similar impedance and mechanical characteristics as the unimpaired human ankle, resulting in a potentially more natural gait with lower risks of secondary injuries. Several powered prostheses that mimic the human ankle’s stiffness have been designed. For example, the powered ankle and knee prosthesis developed by Sup et al. is capable of controlling the mechanical impedance of both the knee and ankle in the sagittal plane [5], [6]. Hitt et al. developed SPARKy, a tendon driven ankle-foot prosthesis capable of generating a net positive work during walking and running [7], [8]. Au et al. developed early prototypes of BiOM, an ankle-foot prosthesis that uses a finite-state machine to identify the state of gait and control the prosthesis to generate the appropriated torques and net positive work during plantarflexion. Although theses prostheses greatly improve the amputees’ gait in the sagittal plane, a better understanding of the ankle, in both the sagittal and frontal planes, may be beneficial in improving lower limb prosthetic devices.

Different studies have explored the 1-degrees of freedom (DOF) ankle impedance in the sagittal plane under no load [9]–[19]; however, this may not be a sufficient representation of the ankle impedance during standing and walking, where the impedance varies during the different phases of each stride [2], [3]. Estimation of the mechanical impedance of the ankle in load
bearing condition has been studied in the sagittal plane [20]–[27] and in the frontal plane [28], [29]. Single DOF ankle movements rarely happen during normal gait, resulting in unique challenges in the control of ankle-foot prosthesis with multiple DOFs [30]. This implies that a comprehensive understanding of the time-varying and task-dependent ankle impedance of the human gait during different types of gait such as turning, traversing slopes, and walking up or down slopes can greatly advance the state of the art in lower extremity assistive devices and robots.

The time-varying impedance of the ankle during walking has been explored by two research groups. Rouse et al. developed a Perturberator, with single active DOF [31]. The Perturberator applied single and isolated perturbations to the ankle in DP at four distinct points of the stance phase during straight walk of unimpaired subjects. The perturbations were applied within 13% up to 63% of the duration of the stance phase, showing great variability in the ankles stiffness (1 N·m/rad/kg at 0.1 s and 4.6 N·m/rad/kg at 0.475 s into the stance phase) and damping (0.012 N·m/rad/kg at 0.1 s and 0.038 N·m/rad/kg at 0.475 s into the stance phase) with an increased ankle stiffness and damping at push-off compared to the heel-strike [2]. A different study was conducted by Lee et al. on the estimation of time-varying mechanical impedance of ankle in both DP and IE, simultaneously [2]. In their study, the subjects used a wearable robot that applied continuous perturbations to the human ankle while the subject walked on a treadmill. The information about the ankle properties during the swing phase, early stance, and toe-off was used to estimate the time-varying impedance of the ankle during those phases using a method developed for identification of physiological systems with time-varying dynamics [32]. Their protocol did not determine the ankle impedance during the mid-stance phase of the gait; however, it showed great variability of the ankle impedance in both DP and IE directions during different phases of walking on a straight path.

This paper presents the design and initial validation of the developed instrumented vibrating platform that is capable of applying torque perturbations to the ankle in two DOFs. The instrumented vibrating platform is the first device, to the best of the authors’ knowledge, which is designed to estimate the ankle impedance in two degrees of freedom in load-bearing conditions. The developed device expands the functionality of the Perturberator developed by Rouse et al. [31] to estimate the time-varying impedance of the ankle in both DP and IE by applying torque perturbations to the ankle in those two DOFs.

The paper describes the hardware and analytical methods for estimation of the impedance of the human ankle in two DOF. It first describes in detail the design and fabrication of the instrumented vibrating platform. Next, it discusses the evaluation of the developed system using an ankle-foot mockup to demonstrate the capability of the system and the developed protocols in identifying the mechanical impedance of a physical system in two DOFs. The analytical methods to calculate the ankle torques and angles based on the information obtained from the force plate and camera system are explained. Additionally, the stochastic identification method that was used for impedance estimation of the ankle-foot mockup based on the ankle angles and torques is presented.

II. INSTRUMENTED VIBRATING PLATFORM DESIGN

The developed vibrating platform (see Fig. 1) consists of two independent modules: 1) actuation module; and 2) the force plate module. The actuation module contains all the power electronics including servo drivers and voice coil linear motors. The force plate module consists of a force plate mounted on a tilting mechanism allowing the force plate to rotate in two DOFs about the longitudinal and lateral axes of the force plate. The force plate module was design to be easily installed in walkways for gait analysis and has a compact height of 0.159 m. This low profile is achieved by separating the actuation and the force plate modules. The power from the actuation module is transferred to the force plate module by four Bowden cables generating torque perturbations.

A. Actuation Module Design

The actuation module (see Fig. 2) is driven by two voice coil linear motors (C) (GVCM-095-089-01, Moticont, USA). The motors are rated at 351 N of force at 10% duty cycle and 111 N continuous force. They have a 63.5 mm stroke length, a coil
resistance of 3 Ω, and coil inductance of 2.1 mH at 1000 Hz. The voice coils motors are powered by a pair of digital servo drivers (D) (510-03, Moticon, USA) capable of delivering up to 40 A at 20 kHz pulse width modulation frequency. The servo drivers are connected to two deep cycle batteries providing 24 V and 114 A·h capacity. The motors run with open-loop current control and receive analog control signals from a DAC (E) (NI USB 6009, National Instrument, USA) connected to a computer. Each motor is connected to two ultraflexible steel cables (F) with 1.6 mm diameter (3423T29, McMaster-Carr, USA). The cables are coated with fluorinated ethylene propylene for its nonstick properties helping to reduce drag and wear in the Bowden cables. Two cables are attached to each end of each motor; therefore, each motor can pull on one cable while releasing the tension on the cable at the opposite side. The Bowden cable housings are attached to flexible carbon fiber springs (G) (8194K56, McMaster-Carr, USA) that are preloaded to assure the cables are always under tension. The flexible carbon fiber springs are mounted to a rail (H), allowing the springs to slide and lock in the same axis as the tension of the cables. This allows for easy adjustment of the position of the motors with respect to the force plate so the force plate becomes horizontal at the mid stroke of the motors. Additionally, the rail allows for preloading of the tension of the steel cables. Fine adjustment of the cable tension and centering the motors (to the mid stroke of the motors) to level the force plate can be obtained by adjusting the Bowden cable housing adjusters (I) (935, Parts Reloaded, USA) located at the carbon fiber springs. Adequate preload in the cables is important as excessive cable tension increases friction inside the Bowden cables, increasing wear and decreasing the available power to the force plate module. Insufficient cable preload results in cable slack during operation. For optimum performance, the tension in the cables should be adjusted and calibrated accordingly before performing the experiments. The voice coil motors generate large magnetic fields when operating, which induce a voltage on the motor’s supporting frame, steel cables, and rail. This voltage, if not insulated from the force plate, generates excessive noise in the force plate readings. To avoid this voltage from reaching the force plate module through the steel cable and the Bowden cable housing, electrical insulators were installed. Two nylon “S” hooks (K) rated at 530 N of workload were attached in line with each cable to electrically insulate the cables from the actuation module. The Bowden cable housing, which are made of steel wrapped in plastics, were electrically insulated using nylon sleeves and washer (J), where the cable housing adjusters (I) connect to the carbon fiber springs (G).

B. Force Plate Module

Fig. 3 shows the force plate module with and without the force plate (L) (9260AA3, Kistler, Switzerland) coupled to the upper frame (Q), respectively. This versatile design allows the force plate to be easily attached to and detached from the upper frame so it can be used in other applications. The force plate is equipped with four piezoelectric three-component force sensors, each located at one of the feet of the force plate (M). The force plate feet are inserted into sleeves (N) located at the upper frame. A tight fit between the feet and the sleeves are assured by rubber O-rings around each foot, assuring the force plate is constrained in the horizontal plane with respect to the upper frame. Neodymium magnets (5679K15, McMaster-Carr, USA) are used at the bottom of each sleeve and are rated at 196 N of puling force. The magnets are bolted to the upper frame and hold the force plate feet inside the sleeves; therefore, hold the force plate to the frame and constrain it from moving in the vertical axis during the device operation. This approach facilitates insertion and removal of the force plate into the upper frame, while constraining any relative motion between the two components without the need of any permanent modification of the force plate.

At the center of the upper frame (Q) and at the center of the lower frame (R), a universal joint (O) constrains both frames from any relative translational and rotational motions about the vertical axis. Additionally, it supports the user’s weight during the experiments. To reduce friction and increase load bearing capacity, a compact universal joint with low friction needle bearings was used (8284K73, McMaster-Carr, USA). The rotation of the universal joint corresponds to the relative motion of the upper frame (constrained to the force plate) and the lower frame (constrained to the ground) imposing a motion to the ankle equivalent to DP and IE when a subject’s foot is on the force plate. Springs (P) maintain the force plate centered in a level position. During the experiments, the human subject and the motors will generate torques at the force plate module. The motors run in open-loop current control with no position control while the person’s ankle will generate the torques required for walking. Both of these torques can cause the force plate to move from its central position. The springs are employed to limit the force plate rotations due to the aforementioned torques and return the force plate to its central position when these torques are removed. In other words, the springs counter the torque generated by the human subjects, while the motors apply perturbation torques to the force plate and the human subjects’ ankle, simultaneously. In an experiment using a human subject with a body mass of 100 kg, the mean of the maximum force plate rotations across 400 unperturbed steps (due to the torques generated by the
The points (S) with both ends attached to the two motors at the same side of the cable segment is twice as long as the other two cable segments are mounted in a similar way. In those two similar segments, points (S and V). By rotating the cables upwards using pulleys, the Bowden cables can mount to the upper frame vertically while avoiding a small radius of turn. This allows for a short overall height of the force plate module, decreased friction, and decreased wear in the Bowden cables. As shown in Fig. 4(b), a pulley is installed at the point of connection (V) where the steel cables connect to the upper frame. The cable comes from one Bowden cable housing, turns upwards 90° at the first pulley (U), turns 180° downwards at the pulley (V), and turns 90° at the second pulley (U) back to another Bowden cable housing. On the other side of the force plate [see Fig. 4(a)], the steel cables come out of one of the Bowden cable housings, turn upwards 90° at the pulleys (U), and connect to the upper frame at one of the points (S). There are three segments of cable, which two of them are mounted in a similar way. In those two similar segments, one end of the cables is attached to a motor and the other end is attached to one of the points (S) in the upper frame. The third cable segment is twice as long as the other two cable segments with both ends attached to the two motors at the same side of the actuation module. The midpoint of this cable, where the force plate is at its central position, is at the pulley (V). The points (V) and (S) are at 0.24 m ($L_X$) from the universal joint in the longitudinal axis of the force plate module, but in the opposite sides of the upper frame. These lengths are the moment arms for DP torque generation. The distance between the points (S) to the universal joint is 0.075 m ($L_Y$) in the lateral axis and are the moment arms for the IE torque generation. It is important to use symmetrical moment arms; otherwise, the cable length would change and become slack as the upper frame rotates away from the central position. However, some change in cable length is still observed. Linear motion of the cables generates rotational motion in the upper frame. This effectively changes the cables’ lengths as the upper frame deviates away from its central position as the motion of the points (V) and (S) are not linear with respect to the lower frame. During the experiments, the angular displacement of the upper frame is small ($<2^\circ$), resulting in small changes in cable length. The preloading of the carbon fiber springs (G) in the actuation module can easily accommodate for the changes in the cables’ lengths; therefore, the cables never become slack.

Initial estimation of the platform design requirements showed that the platform’s stiffness should be large enough to result in its minimal angular displacement due to the user input torque. This is an important parameter for minimizing the variations in the gait pattern. On the other hand, the platform needs to be compliant enough to allow the motors to rotate it. The actuation system should generate torques larger than the maximum torque expected from a human’s ankle to be able to generate perturbation torques to the ankle at all times. The ankle torque for a person weighing 75 kg during walking reaches up to 140 N-m in DP [33]. The mechanism in the force plate module uses the additions and differences of the pulling forces of the motors to generate DP and IE torques, respectively. The upper frame was designed with moment arms ($L_X$) of 0.24 m in DP and 0.075 m in IE ($L_Y$), and the motors (maximum pulling force of 351 N each) generate a combined maximum torque of 168 N-m in DP and 26 N-m in IE. The current setup of the platform uses four springs (P) rated at 12 kN/m and are attached at each corner of the force plate module between the upper and lower frames. The springs are preloaded 0.025 m and generate a rotational stiffness of 10.7 kN-m/rad in DP and 7.2 kN-m/rad in IE. The platform stiffness can be adjusted to different users’ or experiments’ requirements by moving the springs closer or farther away from the universal joint. Moving the springs closer or farther away from the universal joint in the longitudinal axis would decrease or increase the stiffness in DP, respectively. Similarly, moving the springs closer or farther away from the universal joint in the lateral axis would decrease or increase the stiffness in IE, respectively. In addition, different spring stiffness values or a combination of springs can be used to generate the desired stiffness of the platform.

A side view diagram of the force plate module with a foot on the force plate is shown in Fig. 5. The force plate coordinate system (XYZ) was defined with the X-axis pointing forward from the human subject point of view, and the Z-axis pointing upwards. The force plate coordinate system is attached to the force plate, and can be described in the global coordinate system ($X_0Y_0Z_0$) using a rotation matrix obtained from the camera system recorded data. The net torque about the universal joint in the Y-axis at the upper frame of the force plate module ($\tau_{DP,FP}$) can be calculated by balancing the torques applied to the upper frame about the Y-axis

$$
\tau_{DP,FP} = J_Y \dot{\theta}_{DP} = -(F_V - F_{S1} - F_{S2})L_X + (K_{1+2}) \nonumber $$

$$
+ K_{(2+3)}L_X \theta_{DP} + (F_{Z(1+4)} - F_{Z(2+3)})L_X \nonumber $$

$$
+ (F_{X(1+4)} + F_{X(2+3)})L_Z \nonumber $$

(1)

where $J_Y$ is the mass moment of inertia of the upper frame about the Y-axis, $F_V$, $F_{S1}$, and $F_{S2}$ represent the motors force at the points V and S (see Fig. 4). The moment arm $L_Y$ is the constant distance (assuming small angles) from the universal joint to the points V and S in the X-axis of the upper frame. With similar assumption, the moment arm $L_Z$ is the constant...
distance from the universal joint to the points $V$ and $S$ in the $Z$-axis of the upper frame.

For simplicity, the sums of two similar variables on the same side of the platform are presented as one variable with a descriptive subscript. At each corner of the force plate module, there is one spring; therefore, $K_{(1+2)}$ represents the sum of the spring constants in one side of the force plate module in the $X$-axis, $J_{X}$ is the mass moment of inertia of the upper frame, $\theta_{DP}$ is the rotation of the upper frame about the $Y$-axis measured from the two force sensors at each side of the force plate module in the $X$-axis. $F_{Z(1+2)}$ and $F_{Z(3+4)}$ represent the measured forces in the $Z$ and $X$-axis respectively. $J_{\text{ankle}}$, $K_{\text{ankle}}$, and $J_{\text{plate}}$ are vectors containing the ankle's mass moment of inertia, rotational damping, and rotational stiffness in DP and IE. CP is the center of pressure of the foot on top of the force plate. AC is the ankle center of rotation in the force plate coordinate system.

Similarly, the net torque about the $X$-axis of the universal joint at the upper frame of the force plate module $(\tau_{\text{IE, FP}})$ can be calculated by balancing the torques about the $X$-axis of the upper frame. Note that only the motors forces applied at the points $S$ (see Fig. 4) contribute to torque generation in IE since the force applied at the point $V$ is aligned with the $X$-axis resulting in a zero moment arm

$$
\tau_{\text{IE, FP}} = J_{X} \dot{\theta}_{\text{IE}} = -(F_{S1} - F_{S2})L_{X} + (K_{(1+4)} + K_{(2+3)})L_{Y} \dot{\theta}_{\text{IE}} + (F_{Z(1+4)} - F_{Z(2+3)})L_{Z} + (F_{Y(1+4)} + F_{Y(2+3)})L_{Y}
$$

where $J_{X}$ is the mass moment of inertia of the upper frame about the $X$-axis. The moment arm $L_{Y}$ is the constant distance (assuming small angles) from the universal joint to the points $S$ along the $Y$-axis of the upper frame. $K_{(1+4)}$ and $K_{(2+3)}$ represent the sum of the springs constants of the two springs on each side of the force plate module in the $Y$-axis. $\dot{\theta}_{\text{IE}}$ is the rotation of the upper frame measured with a motion capture camera system about the $X$-axis. $F_{Z(1+4)}$ and $F_{Z(2+3)}$ represent the sum of the forces in the $Z$-axis and $F_{Y(1+4)}$ and $F_{Y(2+3)}$ represent the sum of the forces in the $Y$-axis measured from both the force sensors in each side of the force plate module in that direction.

In (1) and (2), the measured values are the force values from the force plate and the angles measured by the motion capture camera system. Solving for $\dot{\theta}_{\text{DP}}$ and $\dot{\theta}_{\text{IE}}$ in (1) and (2), and setting $\dot{\theta}_{\text{DP}}$ and $\dot{\theta}_{\text{IE}}$ to zero (static condition), the maximum rotations based on the stiffness of the platform and the maximum force from the motors are determined as $\pm 0.9^\circ$ in DP and $\pm 0.3^\circ$ in IE. The maximum expected torque is obtained when the motors and the human subject apply a torque in the same direction, which in DP adds up to 308 Nm. This torque would cause an upper frame rotation of $\pm 1.8^\circ$ that is in the linear range of operation of the force plate module.

To calculate the ankle torques in DP and IE $(\tau_{\text{DP}}$ and $\tau_{\text{IE}}$), the following equations can be derived:

$$
\tau_{\text{DP}} = (J_{\text{DP, ankle}} + J_{\text{DP, plate}})\dot{\theta}_{\text{DP}} + B_{\text{DP, ankle}} \ddot{\theta}_{\text{DP}} + K_{\text{DP, ankle}} \dot{\theta}_{\text{DP}} = (F_{Z(1+2)} + F_{Z(3+4)}) (CP_{X} - AC_{X}) + (F_{X(1+2)} + F_{X(3+4)}) (CP_{Z} - AC_{Z})
$$

$$
\tau_{\text{IE}} = (J_{\text{IE, ankle}} + J_{\text{IE, plate}})\dot{\theta}_{\text{IE}} + B_{\text{IE, ankle}} \ddot{\theta}_{\text{IE}} + K_{\text{IE, ankle}} \dot{\theta}_{\text{IE}} = (F_{Z(1+4)} + F_{Z(2+3)}) (CP_{Y} - AC_{Y}) + (F_{Y(1+4)} + F_{Y(2+3)}) (CP_{Z} - AC_{Z})
$$

where $J_{\text{DP, ankle}}$, $B_{\text{DP, ankle}}$, and $K_{\text{DP, ankle}}$ are the ankle’s mass moment of inertia, the ankle’s rotational damping, and the ankle’s rotational stiffness in DP, respectively. $J_{\text{IE, ankle}}$, $B_{\text{IE, ankle}}$, and $K_{\text{IE, ankle}}$ are the ankle’s mass moment of inertia, the ankle’s rotational damping, and the ankle’s rotational stiffness in IE, respectively. Also, $J_{\text{DP, plate}}$ and $J_{\text{IE, plate}}$ are the force plate’s mass moment of inertia in DP and IE, respectively. $CP_{X}$, $CP_{Y}$, and $CP_{Z}$ are the center of pressure of the foot on top of the force plate in the X, Y, and Z axes, respectively, as described in [1]. $AC_{X}$, $AC_{Y}$, and $AC_{Z}$ are the ankle center of rotation in the X, Y, and Z-axes, respectively, that are calculated based on the position and rotations of the foot and shin and will be explored later in this paper.

The effects of the inertia of the force plate $(J_{\text{DP, plate}}$ and $J_{\text{IE, plate}}$) are also present on force measurement. The force plate has a mass of 5.5 kg; however, its mass moment of inertia is unknown. A simple experiment applying random input perturbations to the force plate and recording the output motion was conducted in order to estimate the impedance of the force plate. Therefore, the impedance of the force plate due to its inertia,
C. Instrumented Walkway

The instrumented vibrating platform was used in a walkway as shown in Fig. 6. The walkway allows the placement of the force plate module so the top surface of the force plate is flush with the walking surface of the walkway. This allows human subjects to easily stand or walk on the force plate module. The force plate module records the forces applied to the ankle while the motors apply perturbations and the motion of both the force plate and the ankle are recorded using a motion capture camera system. The walkway is 0.159 m tall (the same as the force plate module), 6 m long, and 1.83 m wide. It is constructed of solid wood to avoid undesired motion that a hollow walkway could produce. The surface is covered with plywood to provide a level and safe ground. The walkway is modular, in the sense that its length and shape can be configured to different types of turning maneuvers, walking on uneven terrain, or slopes. In addition, the force plate module can be replaced by a conventional force plate for standard gait analysis experiments.

The motion capture camera system was used to track the position and rotations of each rigid body of interest during the experiments. The camera system consists of eight Prime 17W Optitrack mounted in a square formation covering a volume of nearly 16 m³ and an area of 12 m². Reflective markers are mounted into polycarbonate plastic rigid bodies developed by the camera system manufacturer to eliminate relative motion of the markers with respect to each other. A polycarbonate plastic rigid body was installed on each rigid body of interest. The camera system emits infrared light, which is reflected back by the reflective markers to the camera sensors. The reflected light is recorded by the camera system at a rate of 300 Hz allowing for the calculation of the position of each marker and subsequent calculation of the position and rotations of each rigid body. The precision of the camera system to track the markers movement during perturbation is important since an inaccurate measurement may have adverse effects on the results of the system identification. During the experiments performed in this paper, the standard deviation of the relative position of two markers on the force plate rigid body during perturbations was 0.6 ± 0.1 μm.

III. System Validation Using a Mockup

A. Mockup Design

To validate the instrumented walkway and vibrating platform capability for estimation of the mechanical impedance of a known system, an experiment with a mockup was conducted (see Fig. 7).

The mockup consisted of a base and an inverted pendulum of 68 kg (667 N) with its center of mass 0.8 m above its center of rotation. The pendulum rotated about a spherical joint, which was placed 0.1 m above the force plate to simulate the distance from the force plate and the ankle in humans. The mockup base was comprised of four horizontal bars spaced in 90° and weighed 1.5 kg. The acceleration of the mockup base generates forces which are recorded by the force plate, so it was designed with weight comparable to the human foot, which has been reported as 1.46% of body weight [34]. Each horizontal bar was connected with a heavy-duty spring rated at 1925 N/m to the main vertical bar of the inverted pendulum. The springs added stiffness to the mockup; however, they were not sufficiently stiff to balance the inverted pendulum. To help balancing the mockup, especially when the platform applied perturbations to the mockup, a supporting spring rated at 3750 N/m was attached vertically to the top of the mockup. The supporting spring was preloaded until normal force on the force plate reached 600 N, indicating that the supporting spring was bearing 67 N out of the 667 N of the mockup’s weight. Note that the force generated by the vertical spring does not add to the stiffness of the mockup when it is vertical, but if the mockup moves away from its vertical position, it will generate a restoring force rotating the mockup back to the vertical position. This restoring torque is an unmeasured input to the system, which may be similar to the...
conditions during experiments on human subjects when they use handrails for balancing.

To record the angular displacement of the mockup base, infrared markers were installed on the mockup base and the inverted pendulum (three markers on each body) that can be regarded as the equivalent of the human foot and shin, respectively.

B. Calculation of the Ankle Center of Rotation

The torque calculation, as shown in (3) and (4), requires the position of the center of rotation of the ankle (AC) in the force plate coordinate system. The ankle center of rotation is the instantaneous center of rotation of the foot with respect to the shin. The point AC can be transformed from the ankle center of rotation in the global coordinate system \( P_{\text{ankle}} \) given the motion capture camera system measurements. To calculate \( P_{\text{ankle}} \), it was assumed that the foot and the shin segments are rigid bodies connected by a spherical joint. This kinematic constraint enforces the existence of a single pivot between the two rigid bodies. This pivot is the ankle center, as shown in Fig. 8. In humans, the ankle center of rotation is not identical to a spherical joint. The human ankle is composed of the talocrural joint and the subtalar joint, and these joints are not aligned with the anatomical axes. The proper calculation of each instantaneous center of rotation is not trivial, and for simplicity, the human ankle can be approximated well as a spherical joint to reduce computational costs, mechanical complexity, and numerical instabilities [35], [36].

The vector \( r_{SA} \) defines the ankle’s center of rotation in the shin coordinate system \( (X_1 Y_1 Z_1) \). Similarly, the vector \( r_{FA} \) defines the ankle’s center of rotation in the foot coordinate system \( (X_2 Y_2 Z_2) \). The vectors \( r_{SA} \) and \( r_{FA} \) can be defined in the global coordinate system using the rotation matrices \( R_{\text{shin}} \) and \( R_{\text{foot}} \), respectively. The rotation matrices \( R_{\text{shin}} \) and \( R_{\text{foot}} \) are found from the motion capture camera system. The global position of the ankle center of rotation \( P_{\text{ankle}} \) at any recorded instance \( i \) is defined in

\[
P_{\text{ankle}} = P_{\text{shin}} + R_{\text{shin}} r_{SA} \tag{5}
\]
\[
P_{\text{ankle}} = P_{\text{foot}} + R_{\text{foot}} r_{FA} \tag{6}
\]

where \( P_{\text{shin}} \) and \( P_{\text{foot}} \) are vectors that define the foot and shin coordinate systems in the global coordinate system. Even though both (5) and (6) define the coordinates of the ankle center of rotation, using the data from the camera system results in a small difference between the two vectors \( \epsilon \) at each instance. This difference can be rewritten into a linear least square problem using the position and orientation of all the recorded samples (N) during the experiment as shown in (7). Where \( \Phi \) is a matrix that contains all the \( R_{\text{shin}} \) and \( R_{\text{foot}} \) matrices, \( \Delta P \) is a vector that contains all the \( P_{\text{shin}} - P_{\text{foot}} \) vectors, and \( \beta \) is a vector that contains the optimal solution vectors \( r_{SA} \) and \( r_{FA} \) that
minimizes the residual error vector $\epsilon$. In the experiments presented in this paper, the data were collected at 300 Hz for 60 s, resulting in $N$ to be equal to 18 000 samples
\[
\begin{pmatrix}
R_{iI}^{shin} & -R_{iI}^{foot} \\
\vdots & \vdots \\
R_{N}^{shin} & -R_{N}^{foot}
\end{pmatrix}
\Phi
\begin{pmatrix}
\tau_{i}^{SA} \\
\vdots \\
\tau_{N}^{SA}
\end{pmatrix}
+
\begin{pmatrix}
P_{iI}^{shin} - P_{iI}^{foot} \\
\vdots \\
P_{N}^{shin} - P_{N}^{foot}
\end{pmatrix}
= \epsilon
\]
(7)

Vectors $\tau_{SA}$ and $\tau_{FA}$ can be calculated by minimizing the residual vector $\epsilon$ using the linear least square method in (7). By substituting $\tau_{SA}$ and $\tau_{FA}$ in either (5) or (6), the ankle center of rotation $P_{ankle}$ can be found in the global coordinate system. In this paper, the average results from both equations (5) and (6) was determined and used for increased accuracy. The point $P_{ankle}$ was later transformed to the force plate coordinate system for determining the point AC to be used in the ankle torque calculations.

C. Mockup Impedance Measurement Using a Single-Variable Stochastic Identification Method

The impedance of a system can be defined as a transfer function $Z(f)$ relating input angles to the output torques at different frequencies $f$. A second order system’s impedance can also be defined by its inertia, damping, and stiffness. The impedance measurement of the mockup consisted of generating uncorrelated pseudorandom torque perturbations at each motor, generating random torque inputs to the mockup in both DP and IE simultaneously. The bandwidth of the perturbations was set to 30 Hz and the sampling rate of the force plate and camera system was set to 300 Hz. A system with torque inputs and motion outputs is defined as a mechanical admittance. Assuming a linear dynamics, the admittance $Y(f)$ is a transfer function relating the input torque $\tau(f)$ and the output rotation of the mockup $\theta(f)$ (8). The inverse of the admittance function is the impedance function $Z(f)$ relating the input angles to the output torques (9)

\[
\theta(f) = Y(f)\tau(f) \quad (8)
\]
\[
Z(f) = Y^{-1}(f). \quad (9)
\]

Multiplying both sides of (8) by $Z(f)$, substituting $Y^{-1}(f)$ to the right side, and solving for the impedance $Z(f)$:

\[
Z(f) = \frac{\tau(f)}{\theta(f)}. \quad (10)
\]

where the values of $\tau(f)$ and $\theta(f)$ are determined from the experimental measurements. The force plate readings include the effects from both the force plate and mockup. As a result, the estimated impedance is a sum of the mockup and force plate impedances. To obtain the mockup impedance, it is necessary to subtract the impedance of the force plate from the estimated combined impedances

\[
Z(f)_{\text{mockup}} = Z(f)_{\text{mockup+force plate}} - Z(f)_{\text{force plate}}. \quad (11)
\]

The impedance of the plate $(Z(f)_{\text{force plate}})$ was obtained from a similar experiment with no mockup on top of the force plate. All the force plate and camera system readings were due to the inertia of the force plate as there is no external spring or damping acting on the force plate. The transfer function relating the input angles and output torques was calculated using a stochastic identification method. MATLAB function $\text{fesimate}$ with a periodic Hamming window and a length of 512 samples with 50% overlap was used. A fast Fourier transform (FFT) of 1024 samples resulted in a frequency resolution of 0.293 Hz. The coherence function was also calculated in MATLAB using the $\text{mscohere}$ function to estimate the linear relationship between the input angles and output torques. The plot of the magnitude, phase, and coherence of the mockup in both axes of the base, equivalent to DP and IE direction of human ankle-foot, are shown in Figs. 9 and 10, respectively. The magnitude of the impedance plot at 0.9 Hz was 227.3 N·m/rad (47.28 dB) for DP and 231.2 N·m/rad (47.28 dB) in IE. The value at 0.9 Hz was chosen as it shows the highest coherence below the break frequency in all tests (before the effects of inertia become dominant in the bode plots) and properly represents the stiffness of the mockup with minimal effect of damping and inertia. Figs. 9 and 10 show the results with and without removing the dynamics of the force plate. It can be seen that the inertia of
the force plate affects the magnitude and phase only above the break frequency where the inertia plays a more dominant role. This effect is more relevant in DP since the inertia of the force plate is higher about its lateral axis.

D. Mockup Impedance Measurement Using a Multivariable Stochastic Identification Method

A similar approach can be used to calculate the impedance of the mockup using a multivariable stochastic identification method as described in Appendix A. This approach takes into the consideration that the input motion in DP may affect the output torque in IE and vice versa due to the coupled dynamics [37]. The plot of the impedance of the mockup using a multivariable approach is shown in Fig. 11 with the effects of the force plate dynamics removed. Insignificant differences were observed when the single-variable and multivariable stochastic system identification methods were compared since the mockup has low mechanical coupling between DP and IE. The average relative magnitude error between the single-variable and multivariable methods averaged across all frequencies in the 0 to 15 Hz was 2.11% in DP and 1.21% in IE, respectively. For the human ankle however, the amount of coupling between DP and IE during standing and walking is unknown, suggesting a multivariable impedance estimation approach to be more plausible. The magnitude of the impedance plot at 0.9 Hz was 229.2 N·m/rad (47.21 dB) for DP and 232.8 N·m/rad (47.34 dB) in IE.

E. Quasi-static Stiffness Measurement

An experiment was performed with the mockup to estimate its stiffness independently of the instrumented platform and camera system (see Fig. 12). In this experiment, the entire mockup was lifted and its weight was supported by only two of the horizontal bars in its base. Underneath each of the two supported horizontal bars, an aluminum cylinder was placed to create a revolute joint allowing the rotation of the base, including the other two horizontal bars, about the supported axis. The experiment was performed by adding 4 weights consecutively to one of the unsupported horizontal bars and measuring the vertical displacements of the bar’s end with respect to the ground. Each weight had a mass of about 1.2 kg, and by consecutively adding the weights, torques of 3.2, 6.3, 9.7, and 13.3 N·m were generated. These torque values were estimated based on the mass of each weight, and the distance from the weight was axis of rotation of the mockup’s base. The weights were chosen such that their imposing torque values to be comparable to the torques used during the experiment with pseudorandom perturbations (described later in the paper). During that experiment, the root mean square of the torque perturbations was 10.26 and 4.71 N·m for DP and IE, respectively. The experiments with weights were conducted twice, once on each axis of rotation of the mockup.

The mockup rotation is defined as the rotation of the vertical bar in inverted pendulum with respect to the base; however, measuring the displacement of the pendulum with respect to the ground was impractical. As a remedy, the inverted pendulum was constrained from motion. In this experiment, a chain was used to constrain the inverted pendulum from rotating in the direction of the applied torque. Therefore, the rotations due to the applied torque could be calculated based on the displacement of the mockup base alone. Using the displacement of the horizontal bar where the weights were installed, the rotations of the mockup’s base were calculated. Using the measured base rotations and the applied torques, MATLAB function polyfit was used to create a first-order polynomial fit (straight line) between
the angles and torques. The `polyfit` function finds the coefficients of a polynomial of degree $n$ that fits the data in a least squares sense. From the first-order fit, the slope (stiffness) was obtained, resulting in a stiffness of 233.1 N·m/rad in one axis of rotation of the mockup (equivalent to DP). The experiments were repeated for the other axis of rotation of the mockup after switching the supported horizontal bars, which resulted in a stiffness of 231.5 N·m/rad (equivalent to IE).

Comparing the results of the quasi-static stiffness measurements using weights and a micrometer and the results of the impedance measurement using the single-variable stochastic identification method at 0.9 Hz, the relative errors were 2.57% and 0.15% for DP and IE, respectively. Comparing the results to the impedance measurement using the multivariable stochastic identification method at 0.9 Hz, the relative errors were 1.68% and 0.54% for DP and IE, respectively.

IV. DISCUSSION

The ankle-foot mockup stiffness was estimated using weights and a micrometer and the results were compared to the estimated impedance magnitude using both single-variable and multivariable stochastic identification methods. Both stochastic identification methods used the pseudorandom torque perturbations generated by the motors that caused angular displacements measured by the motion capture camera system. The stiffness of the mockup was within 2.57% for DP and 0.15% for IE when the results of the quasi-static experiment were compared to the results of the single-variable stochastic identification method. In addition, when the results of the quasi-static experiment were compared to the results of the multivariable stochastic identification method, the relative errors were 1.68% in DP and 0.54% in IE. This showed that the camera system and force plate provided reliable information to calculate the ankle torques, angles, and center of rotation that were used in the impedance estimation. The accuracy of the passive markers, which could be a source of error, showed no significant effect in the results as the stiffness estimated from the experiment without the camera system (using weights and a micrometer) was similar to the stiffness obtained from the impedance estimation using the camera system. In addition, the coherence of the impedance plots was above 0.8 (except around the natural frequencies observed at near 6 and 14 Hz) showing low measurement noise and close to a linear correlation between the torques and angles. Moreover, the results showed that the torque generated by the dynamics of the force plate module frame and springs and the torques generated by the motors were properly decoupled from the force plate measurements and their effects were not present in the impedance estimation. This is known since these elements were not used in the quasi-static experiment, but both experiments resulted in similar stiffness estimations.

Since the mockup consisted of springs and a spherical joint, no coupling was expected between DP and IE. In the impedance measurements at 0.9 Hz, the magnitude of the impedance was estimated as 227.3 N·m/rad (47.13 dB) for DP and 231.2 N·m/rad (47.28 dB) in IE using the single-variable stochastic identification method. Using the multivariable stochastic identification method, the impedance magnitude estimations were 229.2 N·m/rad (47.21 dB) for DP and 232.8 (47.34 dB) N·m/rad in IE. These values corresponded to a relative error of 0.84% for DP 0.69% for IE at this frequency. The mean relative error between the single-variable and multivariable methods averaged across all frequencies within 0–15 Hz was 2.11% in DP and 1.21% in IE, respectively. In the human ankle however, the amount of coupling between DP and IE during standing and walking is unknown, suggesting a multivariable impedance estimation approach to be more suitable.

The mockup had natural frequencies near 6 and 14 Hz in both DP and IE, which are visible by a drop in the magnitude plots due to the low damping ratio of the mockup, a 180° phase change, and the drops in coherence at those frequencies. The coherence values for the estimated impedance of the mockup from both single-variable and multivariable stochastic methods were above 0.8, except around the natural frequencies observed at near 6 and 14 Hz. The drop in coherence is expected since it is usual at natural frequencies. The coherence plots showed a linear relationship between angles and torques in the frequencies of interest. This showed the capability of the system to estimate the impedance of the mechanical system being investigated.

Future research will evaluate the capability of the presented device to estimate the ankle impedance during gait. The research goal is not only the estimation of the ankle impedance in straight walking, but also during turning in arbitrary directions. For this purpose, the presented design has a low profile suitable for being installed into a walkway; therefore, the human subjects do not need to carry any wearable device that might change the dynamic of the gait. The developed device only requires the placement of infrared markers on the subject’s body for tracking the ankle trajectories and rotations. Additionally, the participants are not constrained to walk on a treadmill or merely a straight path.

V. CONCLUSION

This paper described the fabrication and initial evaluation of a 2-DOF instrumented vibrating platform that is aimed for estimation of impedance estimation of the human ankle during standing and walking in arbitrary directions in both the sagittal and frontal planes. The vibrating platform was capable of generating torque perturbations with magnitude similar to the torque generated by human ankle in both DOFs. A force plate and motion capture camera system were used for recording the data necessary for the calculation of the ankle’s torques and angles required for the ankle’s impedance estimation in both dorsiflexion–plantarflexion and inversion–eversion directions. The construction of the device was explained in detail and the method for the calculation of the ankle torques and angles was presented. Additionally, the analytical and numerical approaches for impedance estimation were presented. Validation experiments were developed to evaluate the system capability to estimate the impedance of a physical mockup in two DOFs. The impedance of the mockup at 0.9 Hz, using a multivariable stochastic identification method, was within 1.68% and 0.54% of its quasi-static stiffness in dorsiflexion–plantarflexion and inversion–eversion directions, respectively. The experiment
showed that the developed system was capable of properly estimating the impedance of the system.

REFERENCES

[1] E. M. Ficanha, M. Rastgaard, and K. R. Kaufman, “Ankle mechanics during sidestep cutting implicate need for 2-degrees of freedom powered ankle-foot prostheses,” J. Rehabil. Res. Dev., vol. 52, pp. 97–112, 2015.

[2] E. J. Rouse, L. J. Hargrove, E. J. Perreault, and T. A. Kuiken, “Estimation of human ankle impedance during the stance phase of walking,” IEEE Trans. Neural Syst. Rehabil. Eng., vol. 22, no. 4, pp. 870–878, Jul. 2014.

[3] H. Lee and N. Hogan, “Time-varying ankle mechanical impedance during human locomotion,” IEEE Trans Neural Syst. Rehabil. Eng., vol. 23, no. 5, pp. 755–764, Sep. 2015.

[4] M. S. Orenduff, A. D. Segal, J. S. BERGE, K. C. Flick, D. Spanier, and K. G. Klute, “The kinematics and kinetics of turning: Limb asymmetries associated with walking a circular path,” Gait Posture, vol. 23, pp. 106–111, 2006.

[5] F. Sup, A. Bohara, and M. Goldfarb, “Design and control of a powered transfemoral prosthesis,” Int. J. Robot. Res., vol. 27, pp. 263–273, 2008.

[6] F. Sup, “A powered self-contained knee and ankle prosthesis for normal gait in transfemoral amputees,” Ph.D. dissertation, Dept. Mech. Eng., Vanderbilt Univ., Nashville, TN, USA, 2009.

[7] J. Hitt, J. Merlo, J. Johnston, M. Holgate, A. Bohler, K. Hollander, and T. Sugar, “Bionic running for unilateral transfemoral military amputees,” J. Ortho. Sports Phys. Therapy, vol. 26, pp. 244–252, 1997.

[8] J. K. Hitt, T. G. Sugar, M. Holgate, and R. Bellman, “An active foot-ankle prosthesis with biomechanical energy regeneration,” J. Med. Devices, vol. 4, pp. 011003–1–011003–9, 2010.

[9] J. Harlaar, J. Becher, C. Snijders, and G. Lankhorst, “Passive stiffness characteristics of ankle plantar flexors in hemiplegia,” Clin. Biomech., vol. 15, pp. 261–270, 2000.

[10] S. J. Rydahl and T. G. Hargrove, “Ankle stiffness and tissue compliance in stroke survivors: A validation of myotonometer measurements,” Arch. Phys. Med. Rehabil., vol. 85, pp. 1631–1637, 2004.

[11] A. Lamontagne, F. Malouin, and C. L. Richards, “Viscoelastic behavior of plantar flexor muscle-tendon unit at rest,” J. Ortho. Sports Phys. Therapy, vol. 26, pp. 244–252, 1997.

[12] I. W. Hunter and R. E. Kearney, “Dynamics of human ankle stiffness: Variation with mean ankle torque,” J. Biomech., vol. 15, pp. 742–752, 1982.

[13] R. E. Kearney and I. W. Hunter, “Dynamics of human ankle stiffness: Variation with displacement amplitude,” J. Biomech., vol. 15, pp. 753–756, 1982.

[14] R. E. Kearney and I. W. Hunter, “System identification of stretch reflex dynamics,” Crit. Rev. Biomed. Eng., vol. 18, pp. 55–87, 1990.

[15] P. L. Weiss, R. E. Kearney, and I. W. Hunter, “Position dependence of ankle joint dynamics—I. Passive mechanics,” J. Biomech., vol. 19, pp. 727–735, 1986.

[16] R. E. Kearney, R. B. Stein, and L. Parameswaran, “Identification of intrinsic and reflex contributions to human ankle stiffness dynamics,” IEEE Trans. Biomed. Eng., vol. 44, no. 6, pp. 493–504, Jun. 1997.

[17] R. F. Kirsch and R. E. Kearney, “Identification of time-varying stiffness metrics of the human ankle joint during an imposed movement,” Exp. Brain Res., vol. 114, pp. 71–85, 1997.

[18] M. M. Mirbagheri, R. E. Kearney, and H. Barbeau, “Quantitative, objective measurement of ankle dynamic stiffness and intersubject reliability and intersubject variability,” presented at the 18th Annu. Int. Conf. IEEE Eng. Med. Biology Soc., Amsterdam, The Netherlands, 1996.

[19] T. Sinkjaer, E. Tofli, S. Andreassen, and B. C. Hornemann, “Muscle stiffness in human ankle dorsiflexors: Intrinsic and reflex components,” J. Neurophysiol., vol. 60, pp. 1110–1121, 1988.

[20] M. Palmer, “ Sagittal plane characterization of normal human ankle function across a range of walking gait speeds,” M.S. thesis, Dept. Mech. Eng., Mass. Inst. Technol., Cambridge, MA, USA, 2002.

[21] R. Davis and P. DeLuca, “Gait characterization via dynamic joint stiffness,” Gait Posture, vol. 4, pp. 224–231, 1996.

[22] R. C. Fitzpatrick, J. L. Taylor, and D. I. McCloskey, “Ankle stiffness of standing humans in response to imperceptible perturbation: Reflex and task-dependent components,” J. Physiol., vol. 454, pp. 533–547, 1992.

[23] I. D. Loram and M. Lake, “Direct measurement of human ankle stiffness during quiet standing: The intrinsic mechanical stiffness is insufficient for stability,” J. Physiol., vol. 545, pp. 1041–1053, 2002.

[24] S. Sasagawa, J. Ushiyama, K. Masani, M. Kouzaki, and H. Kanehisa, “Balance control under different passive contributions of the ankle extensors: Quiet standing on inclined surfaces,” Exp. Brain Res., vol. 196, pp. 537–544, 2009.

[25] D. A. Winter, A. E. Patla, S. Rietdyk, and M. G. Ihsan, “Ankle muscle stiffness in the control of balance during quiet standing,” J. Neurophysiol., vol. 85, pp. 2630–2633, 2001.

[26] P. G. Morasso and V. Sanguineti, “Ankle muscle stiffness alone cannot stabilize balance during quiet standing,” J. Neurophysiol., vol. 88, pp. 2157–2162, 2002.

[27] K. Shamaei, G. S. Sawicki, and A. M. Dollar, “Estimation of quasi-stiffness and propulsive work of the human ankle in the stance phase of walking,” PLoS ONE, vol. 8, e59935, pp. 1–12, 2013.

[28] A. Saripalli and S. Wilson, “Dynamic ankle stability and ankle orientation,” presented at the 7th Symp. Footwear Biomechanics Conf., Cleveland, OH, USA, 2005.

[29] S. M. Zinder, K. P. Granata, D. A. Padua, and B. M. Gansneder, “Validity and reliability of a new in vivo ankle stiffness measurement device,” J. Biomech., vol. 40, pp. 463–467, 2007.

[30] A. Arradt, P. Wolf, A. Liu, C. Nester, A. Stacoff, R. Jones, P. Lundgren, and A. Lundberg, “Intrinsic foot kinematics measured in vivo during the stance phase of slow running,” J. Biomech.Eng., vol. 40, pp. 2672–2678, 2007.

[31] E. J. Rouse, L. Hargrove, E. Perreault, M. Peshkin, and T. Kuiken, “Development of a mechatronic platform and validation of methods for estimating ankle stiffness during the stance phase of walking,” J. Biomech. Eng., vol. 135, pp. 10091–10098, 2013.

[32] M. Lortie and R. E. Kearney, “Identification of physiological systems: Estimation of linear time varying dynamics with non-white inputs and noisy outputs,” Med. Biol. Eng. Comput., vol. 39, pp. 381–390, 2001.

[33] J. W. Samuel, K. Au, and H. Herr, “Biomechanical design of a powered ankle-foot prosthesis,” presented at the Int. Conf. Rehabil. Robotics, Noordwijk, The Netherlands, 2007.

[34] C. Clauser, Weight, Volume, and Center of Mass of Segments of the Human Body. Springfield, VA, USA: Nat. Tech. Inf. Service, 1969.

[35] N. Sancisi, V. Parenti-Castelli, B. Baldisserri, C. Belvedere, M. Romagnoli, V. D’Angeli, and A. Leardo, “Validation of an one degree-of-freedom spherical model for kinematics analysis of the human ankle joint,” J. Foot Ankle Res., vol. 5, pp. 1–2, 2012.

[36] N. Sancisi, B. Baldisserri, V. Parenti-Castelli, C. Belvedere, and A. Leardini, “One-degree-of-freedom spherical model for the passive motion of the human ankle joint,” Med. Biol. Eng. Comput., vol. 52, pp. 363–373, 2014.

[37] M. Rastgaard, H. Lee, E. Ficanha, P. Ho, H. Krebs, and N. Hogan, “Multi-directional dynamic mechanical impedance of the human ankle: A key to anthropomorphism in lower extremity assistive robots,” in Neuro-Robotics, vol. 2, P. Artemiadis, Ed. Berlin, Germany: Springer-Verlag, 2014, pp. 157–178.

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