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Validation of Methods for Determining Ankle Stiffness During Walking Using the Perturberator Robot

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Abstract—**Recently developed powered ankle prostheses are capable of providing users with biologically inspired control during walking. However, currently, the appropriate dynamic mechanical properties, or impedance, of the human ankle during walking is unknown. Before trustworthy estimates of the ankle's impedance can be obtained using the Perturberator robot, it must be thoroughly validated. In this study, the sensitivity of standing ankle impedance estimates to foot placement was investigated. Additionally, linear filters that mapped acceleration of the Perturberator motor angle to the forces caused by the robot's intrinsic impedance were determined. Lastly, impedance estimates of a prosthetic foot were obtained at four perturbation timing points during the stance phase of walking and compared to values obtained from an independent measure of prosthetic ankle stiffness. During standing, foot placement had a significant effect on ankle impedance measurements ($p < 0.001$). The linear filters accounted for, on average, 98% of the variance in the forces caused by a perturbation. Lastly, when the impedance of the prosthetic foot was determined during walking, there was 3% error when compared to the stiffness measured by the independent measure at the appropriate timing in stance phase. This work was a preliminary, but important step toward our goal of determining the impedance of the human ankle during walking.**

I. Introduction

The dynamic relationship between the position of a perturbed joint and the response torque is known as joint impedance. This is a fundamentally important property and governs how we interact with our environment and navigate disturbances from our intended motion [1-3]. Impedance has been studied in a variety of joints of the upper and lower-limbs, including the wrist [4], elbow [5, 6], knee [7] and ankle [8-10]. Specifically, measurements of ankle impedance have been determined for a broad array of static tasks, including sitting [11], laying supine [12], and standing [13, 14]. These studies have shown how joint impedance varies with many factors, including mean joint position [15, 16], neural activation [17, 18], perturbation amplitude [9], and applied torque [10]. However, no studies have investigated the impedance of the ankle during dynamic tasks, such as walking or running.

During dynamic tasks, researchers have extensively investigated the torque-angle relationship of the ankle, the slope of which is commonly known as the "quasi-stiffness" [19, 20]. This property has appropriately been used to guide the design of passive prostheses, and investigate the passive rotational spring elements needed to recreate hopping and running tasks [21, 22]. However, the quasi-stiffness does not necessarily provide insight into the true mechanical stiffness or impedance of the ankle during these activities. To determine the impedance of the ankle, it must be displaced from its intended motion, or *perturbed*. To this end, the authors have recently developed a platform robot, termed the Perturberator robot, that was designed to perturb the ankle during walking and measure the response [23]. Once known, the impedance of the ankle during walking could be used to inform the design of biologically inspired control systems for recently developed powered ankle prostheses [24, 25]. Currently, the mechanical characteristics of the healthy human ankle are unknown, so robotic prosthesis control does not necisarily provide natural control for the user. Additionally, such ankle impedance measurements may provide a quantitative metric for analysis of functional spasticity post-stroke [26].

The purpose of this study is to lay the foundation for high quality estimates of ankle impedance during walking obtained by the Perturberator robot. Specifically, we investigate the sensitivity of ankle stiffness measurements while standing to foot placement on the Perturberator robot. Also, we demonstrate the removal of forces caused by the robot's own intrinsic mechanical impedance. Finally, we further validate the impedance estimates by comparing stiffness values of a prosthetic foot estimated during walking with stiffness measurements obtained statically from an independent prosthetic foot mechanical testing machine. In

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this study, stiffness refers to the position dependent component of parametric impedance.

II. Methods

A. Sensitivity to Foot Placement

1) Experimental Protocol

Five healthy subjects, three male and two female, participated in this study. The subject's ages ranged from 24 to 28. Subjects gave written informed consent and the experiment was approved by the Northwestern University Institutional Review Board.

Subjects stood barefoot with both feet on the robot platform with their feet shoulder-width apart. The Perturberator robot is a research device that can be used to apply an angular perturbation to the ankle about its center of rotation and record the reaction forces [23]. The Perturberator robot applied a 0.01 radian (0.5°) dorsiflexion ramp perturbation with length of 75 ms. Each trial consisted of thirty perturbations and the time between each perturbation was drawn randomly from a uniform distribution between 10 and 30 seconds. After every 10 perturbations, subjects were provided a rest period, if desired. Subjects were tested in five foot placement positions (i.e. locations from the center of rotation of the Perturberator robot to the center of rotation of the subject's ankle). The placement positions included 3 cm anterior, 1 cm anterior, neutral, 1 cm posterior and 3 cm posterior. The testing order of the foot placement positions was randomized.

Data acquired included force platform information and motor angle measured from the Perturberator robot, as well as ankle angle relative to shank, all sampled at 1 kHz and acquired from a 16-bit data acquisition card and a personal computer using Matlab (The Mathworks, Natick, MA). The motor angle was output from the Perturberator robot with an angular resolution of approximately 1×10^{-4} radians. The angle of the ankle was determined using a Delsys electrogoniometer (Boston, MA). One end of the electrogoniometer was securely fastened to the shank, while the other end was secured to the side of the foot. The sensor was previously calibrated using a protractor as an independent measure of angle (sensitivity: 1.05 rad/V, with 95% confidence interval: ±0.09 rad/V).

2) Data Analysis

All data were low-pass filtered using a bidirectional fourth order Butterworth filter with a cutoff frequency of 20 Hz and segmented to include 100 ms beginning with the ramp perturbation. The forces caused by the Perturberator's intrinsic impedance were subtracted from the force platform data for each trial, yielding the forces from the subject alone (explained in the proceeding sections). Ankle torque was determined by resolving the ground reaction force to the equivalent torque at the ankle joint [27]. The nominal torque—the torque during quiet standing immediately before the perturbation—was removed, leaving only the torque component caused by the perturbation.

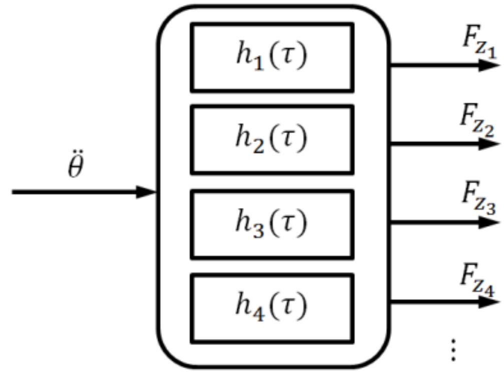

Fig. 1: A series of parallel linear filters are shown mapping acceleration of the Perturberator robot's motor angle to the forces from the force platform. Note, z-axis (vertical) forces shown, but analysis used for all axes.

A second order parametric model was used to identify the impedance of the ankle

$$T_a = I_{tot} \cdot \ddot{\theta} + b_a \cdot \dot{\theta} + k_a \cdot \theta , \tag{1}$$

where T_a is the reaction torque response to the perturbation, I_{tot} is the total inertia of the foot and other coupled body segments; and b_a and k_a are the damping and stiffness coefficients of impedance, respectively; finally, θ is the angular displacement of the ankle. The derivatives were computed numerically in Matlab [28]. The impedance parameters were identified by the least squares estimation method, over the 100 ms window [29]. Variance accounted for (VAF) was used to quantify the agreement of the model with the experimental results. A one-way, repeated measures analysis of variance (ANOVA) was used to determine if foot placement had a significant effect on the stiffness component of impedance and post-hoc comparisons with a Bonferroni correction factor were conducted to analyze differences between placement conditions.

B. Removal of the Perturberator Robot's Intrinsic Impedance

During each perturbation, the swing arm of the Perturberator robot rotates. This causes force transients on the embedded Kistler 9260AA3 portable force platform (Winterthur, Switzerland). In order to obtain force measurements caused by the subject alone, a method was developed for removal of the forces caused by the Perturberator robot's intrinsic inertial impedance.

A single input, multi-output non-parametric model was used to map the acceleration of the Perturberator motor angle to the forces on each channel. A correlation based approach was used to estimate the series of parallel linear filters (Fig. 1) [29]. The estimated filters were causal and had a length of 200 lags. This length was chosen because it adequately captured the response dynamics of the estimated linear filters.

The Perturberator robot's motor angle and force platform data were acquired while the robot made a series of ten, 0.01 radian (0.5°) perturbations with no subject present. Identical data processing techniques were used as when the standing data were analyzed and the data were averaged across trials. A single set of filters was obtained and used to estimate the forces for comparison. Similarly, the VAF was used to quantify model agreement.

Fig. 2: Pneumatic walking brace with prosthetic foot mounted below.

C. Validation of Impedance Estimation during Walking

1) Experimental Protocol

A subject wore custom modified Aircast (Vista, CA) pneumatic walking braces with Trulife (Poulsbo, WA) prosthetic feet (model: SLF165-22-R-H7) mounted underneath (Fig. 2). This permitted the able-bodied subject to walk using the prosthetic feet.

The subject walked on a walkway that included the recessed Perturberator robot (Fig. 3) and a 0.035 radian (2°) dorsiflexion or plantarflexion perturbation was randomly applied to the right foot with a probability of 50%. The duration of the ramp portion of the perturbation was 75 ms. Perturbations were applied randomly at four perturbation timing points during stance phase: 100, 225, 350 and 475 ms following heel strike. One hundred trials were recorded at each perturbation timing points of stance phase and after every 40 perturbation trials, the subject was encouraged to rest. Since there was a 50% probability of a perturbation, there were approximately 400 trials where no perturbation

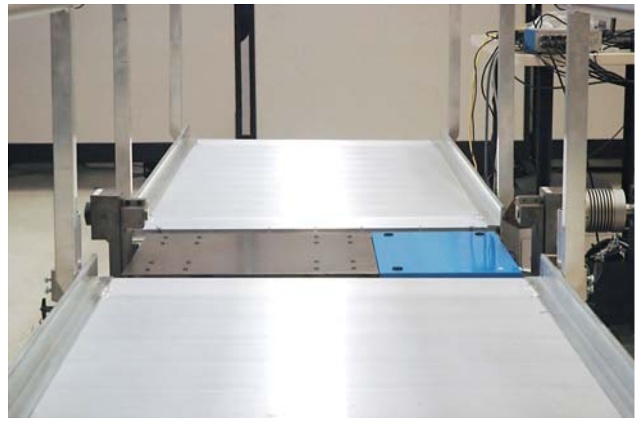

Fig. 3: Perturberator robot shown recessed into walkway. Total length of the walkway was approximately 5.25 meters.

occurred. The data acquired included force platform information, Perturberator motor angle, and prosthetic ankle angle obtained from the Delsys electrogoniometer. All data were sampled at 1 kHz from the 16-bit data acquisition card.

2) Data Analysis

Data were processed using identical techniques to those implemented during the standing experiment. The force and torque were resolved to an estimated location of the ankle center of rotation of the prosthetic foot. Similarly, (1) was used as a parametric representation of the ankle's impedance. However, unlike standing, there were non-zero torque and angle profiles that resulted from walking (Fig. 4). In order to remove this information, a bootstrapping technique was used. A random selection of 80% of the perturbed trials were averaged and subtracted from the average of the non-perturbed trials. If any residual torque and angle remained after the subtraction, it was removed such that both began with zero. Ideally, this technique yielded the torque and angle information that resulted from the perturbation only (Fig. 5). The bootstrapping average-

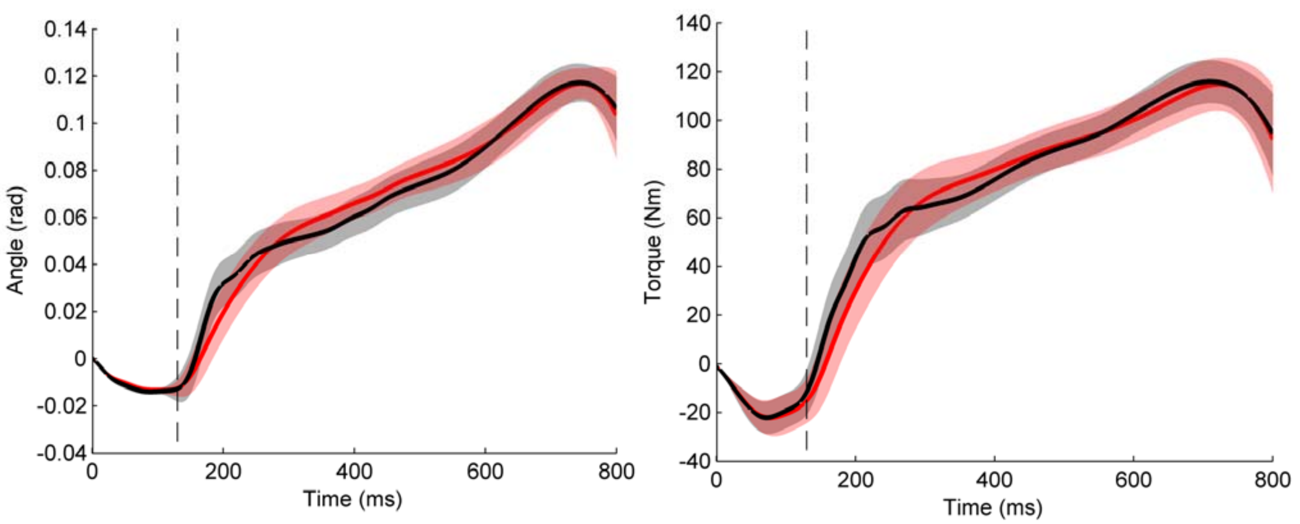

Fig. 4: Prosthetic ankle angle (left) and ankle torque (right). Each is shown with the non-perturbed trials in red and perturbed trials in black (standard deviations in translucent). Data shown are for the earliest perturbation timing points and the perturbation onset is shown as the vertical dotted line.

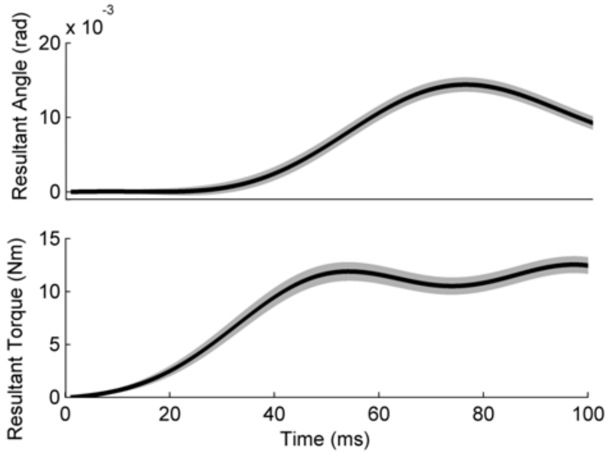

Fig. 5: Resultant angle (top) and torque (bottom) plots shown after bootstrapping and subtraction. Average shown in bold with standard deviation in translucent.

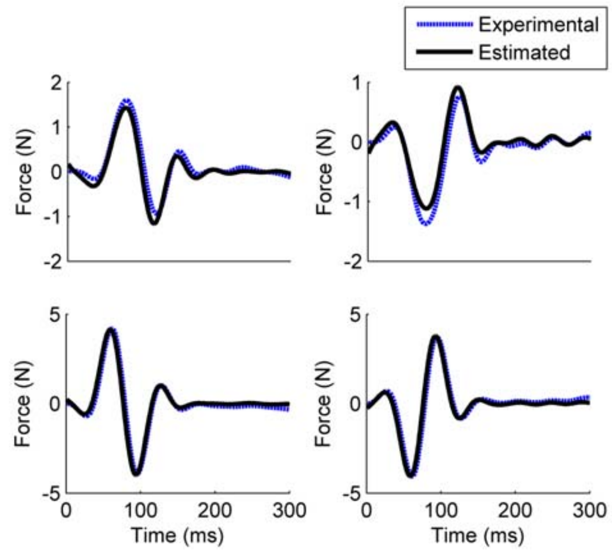

Fig. 7: Estimated (black) and experimental (blue) z-axis forces shown for a representative perturbation trial using the Perturberator robot. The force sensors closest to the motor had the greatest force magnitude.

then-subtract technique was repeated 100 times to achieve a distribution for the impedance parameters, each iteration differing in the specific trials included in the perturbation averages. This technique was used to estimate the impedance characteristics at each perturbation timing point of stance phase. VAF was used to quantify model agreement. Finally, using the center of pressure information calculated from force platform measurements, the distance from the center of rotation of the Perturberator robot to the approximate center of rotation of the prosthetic ankle was determined. This was calculated by measuring the distance from the heel of the prosthetic foot to the estimated center of rotation of the ankle.

3) Independent Measure of Prosthetic Foot Stiffness

In order to compare the estimates of prosthetic foot impedance obtained by the Perturberator robot, an independent comparison was needed. Therefore, the stiffness component of impedance of the prosthetic foot was measured using a testing machine that conformed to ISO 10328 standards for prosthetic forefoot loading. The foot was secured in the machine and the forefoot was deflected 1.27 cm over 10 seconds. A plate was added between the hydraulic actuator and the prosthetic foot shell to prevent local foam deformation. Five trials were recorded and the geometry of the setup was used to convert from linear measurements to angular displacement and torque. Linear regression was used to determine the slope of the torque-angle curve, yielding an estimate for the stiffness component of impedance and the coefficient of determination was used to demonstrate goodness of fit.

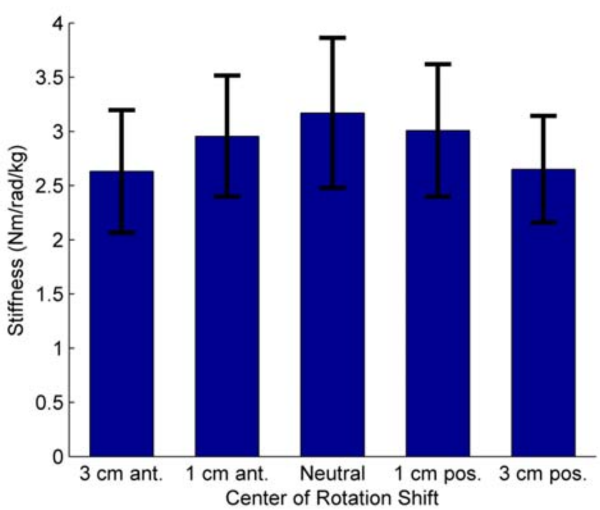

Fig. 6: Ankle stiffness component of impedance shown averaged across subjects. Note that stiffness decreases with foot displacement positions bilaterally.

The distance between the hydraulic actuator and the prosthetic foot's center of rotation was used to determine the appropriate timing comparison during the walking validation experiment. The timing comparison was determined by using the time from heel strike that the center of pressure was equal to the actuator distance during the prosthetic loading (6.3 cm).

III. Results

A. Sensitivity to Foot Placement

The stiffness component of impedance was determined for five subjects while standing at foot placement positions ranging from 3 cm anterior to 3 cm posterior. The stiffness values were averaged across subjects and ranged from 2.6 to 3.1 Nm/rad/kg. The results are shown in Fig. 6. Stiffness decreased with movement of the foot placement position in either direction. The mean percent error was 6% and 17% when displaced 1 cm and 3 cm, respectively. The VAF for the foot placements conditions was very high, consistently above 90%. This indicates a good fit between the model predicted torques and actual torques. The one-way ANOVA showed that foot placement position had a significant effect

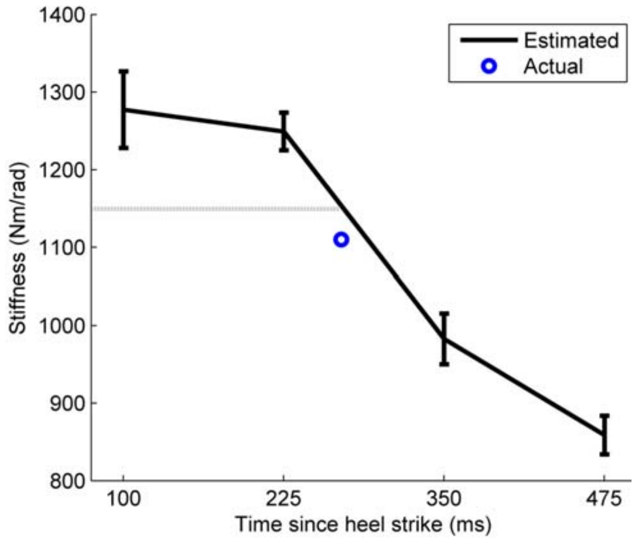

Fig. 8: Prosthetic ankle stiffness measurements determined during walking. The average estimate from the prosthetic foot testing machine is the blue circle when displayed at the appropriate timing (to be compared with the approximate estimated stiffness, denoted by the gray line).

on the stiffness component of ankle impedance ($p < 0.001$). The stiffness estimated in the neutral position was only statistically different than the 3 cm displacement conditions ($p < 0.05$).

B. Removal of the Perturberator Robot's Intrinsic Impedance

Linear filters were estimated that mapped the acceleration of the Perturberator robot's motor to force transients induced in the force platform. The average VAF for the estimated forces was 97.7% with a standard deviation of 2.1%. Exemplary experimental and estimated vertical axis forces are shown in Fig. 7.

C. Validation of Impedance Estimation during Walking

Impedance values for the prosthetic foot were estimated using the Perturberator robot during the stance phase of walking. The stiffness component of impedance (Fig. 8) was determined to range from 1277 Nm/rad to 860 Nm/rad. On average, the VAF was 99.3% with a standard deviation of 0.3%. The stiffness measured from the testing machine ranged from 1099 Nm/rad to 1119 Nm/rad, with a mean of 1110 Nm/rad. The coefficient of determination was consistently above 0.999. When these data were compared to the approximated stiffness estimate during walking (Fig. 8, gray line), there was 3% error in the estimate.

Lastly, the average distance from the center of rotation of the Perturberator robot to the center of rotation of the prosthetic ankle was 1.3 cm (in the anterior-posterior direction), with a standard deviation of 1.5 cm.

IV. Discussion

The goal of this study was to provide a solid foundation for the use of the Perturberator robot in determining human ankle impedance during walking. We initially investigated the sensitivity of the impedance measurements to foot placement during standing. Additionally, we demonstrated a method for removing the transient forces that result from the robot's intrinsic impedance. Lastly, we validated a method for determining the impedance of a prosthetic ankle during the stance phase of walking and compared it to an independent measure of prosthetic foot stiffness.

When foot placement position was varied, there was a significant effect on the impedance estimates. When center of rotation was within 1 cm the mean error was 6%, which increased linearly to 17% when displacement was increased to 3 cm. This suggests that when using the Perturberator robot to estimate ankle impedance in standing or walking paradigms, foot placement should be an experimental consideration; however, the impedance estimates are relatively insensitive to foot placement. This is highlighted by the average placement error during the walking experiment (1.3 cm). The percent error in the walking experiment not only agrees with the general magnitude and direction of error associated with the center of rotation being misaligned (3% error vs. 6% error), but also suggests that more sophisticated controlling of foot placement is unnecessary, if these error magnitudes are acceptable.

In order to provide accurate estimates of ankle impedance, the transient forces that resulted from the perturbation must be removed. The estimated linear filters provided a very accurate representation of the forces, accounting for, on average, approximately 98% of the variance (Fig. 7). Subsequent measurements using the Perturberator robot should have forces arising from the filtered acceleration profile subtracted from each force channel. Essentially, this procedure identified the inertial impedance of the Perturberator robot. This inertial impedance superimposed additional force information on the measurements being used to determine ankle impedance. Since the Perturberator robot's interaction is not in parallel with the system being identified the estimated forces may be removed with subtraction. If the perturbation device is in parallel, as in active orthosis-type designs that span the ankle, the device should have very low intrinsic impedance to ensure high quality estimates of ankle impedance.

When the stiffness determined from the prosthetic foot testing machine was compared to the stiffness estimate during walking at the appropriate time, there was a 3% difference in stiffness values. The low error percentage indicates that the Perturberator robot and associated methods provide accurate estimates of stiffness. Further evidence could be obtained by varying the distance of the hydraulic actuator from the prosthetic foot in the testing machine. This would effectively provide stiffness estimates at other timing points during stance phase, further strengthening the evidence for the estimates.

The Perturberator robot was designed to perturb the human ankle during the stance phase of walking. This is distinctly different than previous studies that have investigated the quasi-stiffness of the ankle, which does not require a perturbation [19]. Since the prosthetic foot tested in this study is passive, its impedance can be determined without a perturbation. The slope was obtained for the average ankle torque vs. the average ankle angle (Fig. 4), yielding a stiffness value of 1050 Nm/rad, within 5% of the value determined from the prosthetic foot testing machine.

The estimates of prosthetic foot stiffness from the Perturberator robot had extremely high VAF (above 99%). This was likely caused by the agreement between the model and the prosthetic foot behavior. From the high coefficient of determination, we can conclude that a linear stiffness model fits the quasi-static behavior of the foot quite well, and a similarly high VAF is expected. To assess the sensitivity of the VAF metric to the impedance parameters, the model stiffness parameter was decreased computationally by 25%. This alteration caused the VAF to be reduced, on average, by 70%.

In conclusion, a validated metric for the determination of ankle impedance during walking was introduced. The stiffness estimates were shown to be relatively insensitive to foot placement (~6% error/cm). The estimated stiffness during walking was within 3% of the independently measured stiffness, when compared at the appropriate timing. Future work will focus on the estimation of the impedance of the human ankle during the stance phase of walking. Following the determination of the impedance parameters during walking, the values will be used to inform the design of biologically inspired controllers for powered prosthetic ankles.

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REFERENCES

[1] Takahashi, C.D., Scheidt, R.A., and Reinkensmeyer, D.J.: 'Impedance control and internal model formation when reaching in a randomly varying dynamical environment', J. Neurophysiol., 2001, 86, (2), pp. 1047-1051

[2] Burdet, E., Osu, R., Franklin, D.W., Milner, T.E., and Kawato, M.: 'The central nervous system stabilizes unstable dynamics by learning optimal impedance', Nature, 2001, 414, (6862), pp. 446-449

[3] Scheidt, R.A., Dingwell, J.B., and Mussa-Ivaldi, F.A.: 'Learning to move amid uncertainty', J. Neurophysiol., 2001, 86, (2), pp. 971-985

[4] Lakie, M., Walsh, E.G., and Wright, G.W.: 'Resonance at the wrist demonstrated by the use of a torque motor: an instrumental analysis of muscle tone in man', J Physiol, 1984, 353, pp. 265-285

[5] Popescu, F., Hidler, J.M., and Rymer, W.Z.: 'Elbow impedance during goal-directed movements', Exp. Brain Res., 2003, 152, (1), pp. 17-28

[6] Selen, L.P., Beek, P.J., and van Dieen, J.H.: 'Impedance is modulated to meet accuracy demands during goal-directed arm movements', Exp. Brain Res., 2006, 172, (1), pp. 129-138

[7] Zhang, L.Q., Nuber, G., Butler, J., Bowen, M., and Rymer, W.Z.: 'In vivo human knee joint dynamic properties as functions of muscle contraction and joint position', J. Biomech., 1998, 31, (1), pp. 71-76

[8] Gottlieb, G.L., and Agarwal, G.C.: 'Dependence of human ankle compliance on joint angle', J. Biomech., 1978, 11, (4), pp. 177-181

[9] Kearney, R.E., and Hunter, I.W.: 'Dynamics of human ankle stiffness: variation with displacement amplitude', J. Biomech., 1982, 15, (10), pp. 753-756

[10] Hunter, I.W., and Kearney, R.E.: 'Dynamics of human ankle stiffness: variation with mean ankle torque', J. Biomech., 1982, 15, (10), pp. 747-752

[11] Sinkjaer, T., Toft, E., Andreassen, S., and Hornemann, B.C.: 'Muscle stiffness in human ankle dorsiflexors: intrinsic and reflex components', J. Neurophysiol., 1988, 60, (3), pp. 1110-1121

[12] Kearney, R.E., Stein, R.B., and Parameswaran, L.: 'Identification of intrinsic and reflex contributions to human ankle stiffness dynamics', IEEE Trans. Biomed. Eng., 1997, 44, (6), pp. 493-504

[13] Casadio, M., Morasso, P.G., and Sanguineti, V.: 'Direct measurement of ankle stiffness during quiet standing: implications for control modelling and clinical application', Gait Posture, 2005, 21, (4), pp. 410-424

[14] Loram, I.D., and Lakie, M.: 'Direct measurement of human ankle stiffness during quiet standing: the intrinsic mechanical stiffness is insufficient for stability', The Journal of Physiology, 2002, 545, (3), pp. 1041-1053

[15] Weiss, P.L., Kearney, R.E., and Hunter, I.W.: 'Position dependence of stretch reflex dynamics at the human ankle', Exp. Brain Res., 1986, 63, (1), pp. 49-59

[16] Weiss, P.L., Kearney, R.E., and Hunter, I.W.: 'Position dependence of ankle joint dynamics--I. Passive mechanics', J. Biomech., 1986, 19, (9), pp. 727-735

[17] Weiss, P.L., Hunter, I.W., and Kearney, R.E.: 'Human ankle joint stiffness over the full range of muscle activation levels', J. Biomech., 1988, 21, (7), pp. 539-544

[18] Mirbagheri, M.M., Barbeau, H., and Kearney, R.E.: 'Intrinsic and reflex contributions to human ankle stiffness: variation with activation level and position', Exp. Brain Res., 2000, 135, (4), pp. 423-436

[19] Hansen, A.H., Childress, D.S., Miff, S.C., Gard, S.A., and Mesplay, K.P.: 'The human ankle during walking: implications for design of biomimetic ankle prostheses', J. Biomech., 2004, 37, (10), pp. 1467-1474

[20] Latash, M.L., and Zatsiorsky, V.M.: 'Joint stiffness: Myth or reality?', Human Movement Science, 1993, 12, pp. 653-692

[21] Farley, C.T., and Morgenroth, D.C.: 'Leg stiffness primarily depends on ankle stiffness during human hopping', J. Biomech., 1999, 32, (3), pp. 267-273

[22] Gunther, M., and Blickhan, R.: 'Joint stiffness of the ankle and the knee in running.', Journal of Biomechanics, 2002, 35, (11), pp. 1459-1474

[23] Rouse, E.J., Hargrove, L.J., Peshkin, M.A., and Kuiken, T.A.: 'Design and validation of a platform robot for determination of ankle impedance during ambulation', Proceedings of the International Conference of the IEEE Engineering in Medicine and Biology Society, 2011

[24] Au, S., and Herr, H.: 'On the design of a powered ankle-foot prosthesis. The Importance of series and parallel motor elasticity', IEEE Robotics and Automation Magazine, 2008, 15, (3), pp. 52-59

[25] Au, S.K., Weber, J., and Herr, H.: 'Biomechanical Design of a Powered Ankle-Foot Prosthesis', Proceedings of 2007 IEEE 10th International Conference on Rehabilitation Robotics, 2007, pp. 298-303

[26] Dietz, V., and Sinkjaer, T.: 'Spastic movement disorder: impaired reflex function and altered muscle mechanics', Lancet Neurol, 2007, 6, (8), pp. 725-733

[27] Rouse, E.J., Hargrove, L.J., Perreault, E.J., and Kuiken, T.A.: 'Estimation of human ankle impedance during walking using the Perturberator robot', Proceedings of the IEEE International Conference on Biomedical Robotics and Biomechatronics, 2012, In press.

[28] Scheid, F.J.: 'Schaum's outline of theory and problems of numerical analysis' (Schaum's Outline Series, 1989. 1989)

[29] Ljung, L.: 'System identification' (Wiley Online Library, 1999. 1999)