Comparison of Ankle Torque Reporting Methods for a Robotic Prosthesis

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1 Introduction

The last decade has seen the transition of powered prosthetic devices from academic research into the commercial marketplace. The iWalk BiOM is a powered prosthetic device that adds power to walking gait cycle. The stock BiOM control algorithms use equations, not based upon muscle theory, to control the ankle torque output. We utilize the BiOM as a research platform in order to test a neuromuscular control algorithm based on NAU's Winding Filament Hypothesis (WFH) [1].

The BiOM control algorithm was designed for level walking. We have implemented the WFH into a control algorithm that also is targeted for level walking. However, we also wanted to adapt the hypothesis-based control system for uneven terrain. Such uneven terrain testing scenarios would include stair ascent/descent and progression through soft or variable-level surfaces. We used the robotic platform to conduct level walking experiments for an analysis of ankle torques using both algorithms. The BiOM hardware and traditional inverse dynamic methods were used to collect the torque data during level walking experiments via a force plate and camera [2]. To conduct uneven terrain experiments, using traditional equipment as force plates and cameras becomes difficult. We used the BiOM hardware data stream and inverse dynamics to determine if the BiOM hardware was sufficiently consistent to use without the need of traditional equipment. One challenge to the experimenters was the recognition that the aforementioned observational techniques applied to intact human limb position and joints are not entirely applicable to robotic limb replacements. For example, the "ball of the foot" in an intact lower leg does not exist precisely in the same positional context in a prosthesis with a composite spring foot [3]. Thus, the techniques we used are adapted to the prosthetic equivalent for the lower leg.

2 Methods

The BiOM hardware data stream was obtained wirelessly from the robotic ankle and stored on a computer. Subjects were instructed to walk down a 10 meter runway, in which a forceplate (AMTI BP400600-1000) was embedded. A camera (Vision Research V-series Phantom v5.1), was placed perpendicular to the runway so that the stance phase of the subject could be captured. Markers were placed on the subjects limbs as in Figure 1 [3] to track the leg movements. The force plate reported the forces (F_x, F_y, F_z) and center of pressure (D_x, D_y) of the subject during the stance phase of the stride cycle. The camera operated at 500 frames per second and captured the subject walking in a 0.5m subsection of the runway. The subject walked at prescribed speeds (1.0, 1.25 and 1.5 m/s) and struck the force plate with the prosthesis. Only the fast walking speed of 1.5 m/s will be reported in this paper, for purpose of brevity. The videos were used to track the markers and make kinematic measurements. We used a custom DLTdv5 MatLab code to digitize the video to find the position of each marker and then numerically integrated into velocities and accelerations.

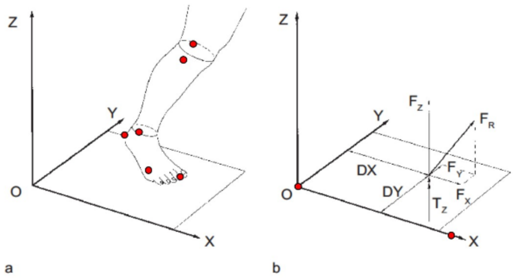

Figure 1. a) Marker placement and b) vector illustration for inverse dynamic calclation of ankle torque [2].

Using the data from the force plate and the high speed camera, Equation 1 was used to calculate the ankle torque.

$$M_{az} = (r_{dy}-r_{py})F_x+(r_{dx}-r_{px})Fy+r_{py}ma_x+r_{px}m(g-a_y)+I_x\alpha_z \quad (1)$$

F_y and F_x are the components of the ground reaction forces measured by the force plate. Components of the proximal (r_{px} , r_{px}) and distal (r_{dx} ,r_{dx}) vectors locate the position of the foot's center of mass. The foot's center of mass acceleration is represented by a_x and a_y. The foot's moment of inertia, mass, and angular acceleration, are represented by I_z, m, and α_z , respectively [3].

3 Results

The torque calculations from inverse dynamics and the BiOM wirelessly output "direct" torque data stream were compared to evaluate the consistency of the BiOM hardware. Due to the dimensional limitations of the facility where the experiments were conducted, the equipment location allowed the researchers to capture only the stance phase (60% of the gait cycle) during each trial.

In the first set of experiments, the BiOM was controlled by the standard "stock" control software. The results were compiled from two methods—either via the inverse dynamic calculations or from the BiOM 'direct' data stream. The data for the experiments were collected at the fast walking speed and 7 trials were averaged. The upper and lower bounds for two standard errors of the mean are illustrated for each torque reporting method. In Figure 2 the stock direct fast speed torque peaks at -45.7 Nm (12% into stance) during heel strike and peaks at 144 Nm (88% into stance) during powered plantar flexion. The Stock inverse dynamics shows that the fast speed torque peaks at -40.7 Nm (19% into stance) during heel strike and 100Nm (81% into stance) during powered plantar flexion.

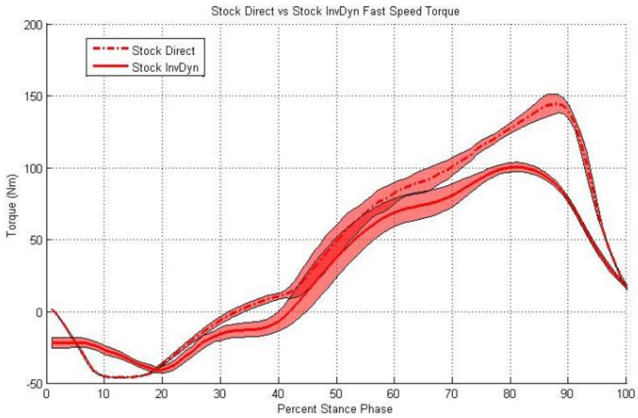

Figure 2. The BiOM stance phase torque, controlled via the stock (unmodified) software.

The second set of experiments were performed where the BiOM was controlled by the WFH-based control software. The two methods were again used to report torque results for the same experiments. Figure 3 illustrates the results. The direct data stream reported the fast speed torque peaks at -44 Nm (14% into stance) during heel strike and peaks at 139Nm (88% into stance) during powered plantar flexion. The inverse dynamics calculations showed that the fast speed torque peaks at -52 Nm (14% into stance) during heel strike and 91Nm (82% into stance) during powered plantar flexion.

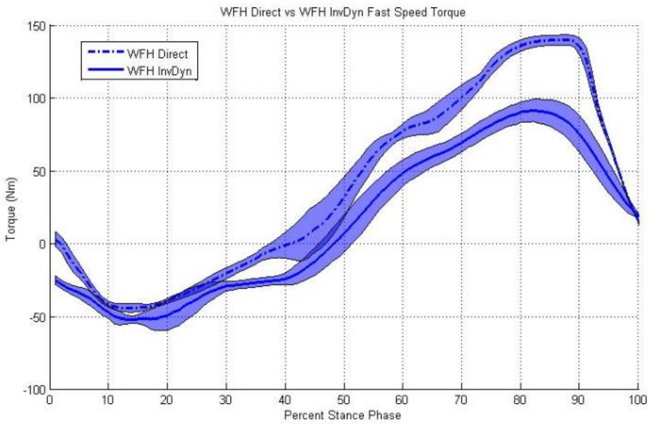

Figure 3. The BiOM stance phase torque, controlled via the WFH-based (modified) software.

4 Discussion

The WFH-controlled BiOM reported similar torque progression with similar peak values when compared to the BiOM that was controlled through its stock control software and for the same torque reporting methods. However, the two torque reporting methods result in similar-shaped curves, but with discrepancies in peak values and their stance-phase locations.

The difference in peak torque reporting were very consistent; in both experiments, the difference in the stance location for the heel strike peak torque was zero to 7%. For the peak torque during powered plantar flexion, the difference was 6% to 7%. The WFH-based control experiments resulted in smoother build-up to the peak torque situations, thus resulting in the two alternative torque reporting methods to be more aligned throughout the stance progression than for the BiOM stock controlled experiments.

5 Conclusion

The direct data stream method for reporting ankle torque did not produce the same magnitudes as that from the inverse dynamics method for throughout the entire stance phase. The most likely reason for the discrepancies is due to the use of a single camera at close range. Typical visual data collection methods for gait analysis uses multiple, high-speed cameras [4]. We could only use one during the experiments, due to limited resources available at the time.

Nevertheless, the two alternative torque reporting methods show similar curvature and variance of the data throughout the stance phase. The magnitude discrepancy at the powered plantar flexion for the two reporting methods is approximately 25% for the entire report torque range (~-50Nm to ~150Nm). These first experiments show that magnitude reporting from either method cannot be reliably reported based solely on these experiments for the powered plantar flexion phase. The direct data stream torque data could be used for comparative studies between subjects or control systems, given the consistency of the stream results. Planning is underway to conduct further testing with more cameras in an improved facility, to improve the alignment of the inverse dynamics reported torque magnitudes with that of the BiOM direct data stream.

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