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Problems associated with the use of inverse dynamics in prosthetic applications: An example using a polycentric prosthetic knee

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SUMMARY

Inverse dynamic calculations are often performed in the analysis of human ambulation. However, assumptions are made in the construction of the models that may not be valid for some prosthetic componentry, specifically polycentric knees used by transfemoral amputees. Data were collected for one subject using a polycentric knee. Moments and forces were calculated when ignoring the extra linkages of the knee and when including the extra linkages of the knee. Significant differences were seen in the hip moment and power between the two cases and indicate that care must be taken when comparing componentry that violates the basic assumptions of the rigid link model.

KEYWORDS: Prosthetics; Inverse dynamics; Polycentric knee.

I. INTRODUCTION

Equations of motion for mechanical systems can be solved in two ways. Forward solutions calculate the trajectories of motion given the forces and torques as input. Inverse solutions calculate the forces and moments given the desired motion. Inverse dynamics are successfully applied in robotics to calculate motor forces and torques, since the characteristics of the system are well defined (masses and inertias are well known, joints are well defined, etc). This technique is also applied to the study of human movement; a common model defines each limb as a rigid link with hinge or ball-and-socket joints. This model is then used to calculate various descriptors of the movement, including joint angles and segmental positions. Through inverse dynamics, this model is also used to calculate the moments or torques that would be required at each joint to produce the observed motion. In human gait studies, powers are also calculated. These powers are defined as the product of the joint moment and the joint angular velocity.

Dynamic equations can be defined in multiple ways. However, standard gait analysis programs included with motion capture systems (e.g. Orthotrack Motion Analysis and Bodybuilder Vicon) typically do not generate the general equations of motion for the entire human body. Instead, they use D'Alembert's principle of equilibrium of bodies in motion. With this type of calculation, a free-body diagram is created for each segment and the internal forces and moments at the joints are calculated by starting the calculations at a linkage where the end forces and moments are known. Bresler and Frankel first performed this kind of computation on human gait in 1950.[1] The positions of the segments were determined using cinematic film and the velocity and accelerations of the segments were then calculated with numeric differentiation. The ground reaction forces and center of pressure of the foot interacting with the floor were measured using a platform instrumented with strain gages. Given mass parameters (from anthropomorphic table data), the internal forces and moments can then be calculated from the individual free-body diagrams for each segment by beginning calculations at the terminal segment (the foot) and using the calculated values as input for the next proximal segment.

With the inverse dynamics calculations applied to human locomotion, various assumptions are made in the development of the model. These are listed by Winter as follows: (i) Each segment has a fixed mass located as a point mass at its center of mass; (ii) The location of each segment's center of mass remains fixed during the movement; (iii) The joints are considered to be hinge (or ball- and -socket) joints; (iv) The mass moment of inertia of each segment about its mass center (or about either proximal or distal joints) is constant during the movement; and, (v) The length of each segment remains constant during the movement.[2] These may be reasonable approximations for the human body, though the foot and knee are not exactly hinge joints and masses may move with muscle contraction. Errors in calculation may be related to assumptions of mass, segment length, joint assumptions, or a combination of these.

Joint assumptions may introduce errors when considering human joints and prosthetic components. Much has been written describing the polycentric movement of the ankle, knee and hip. For example, more precise models of the knee have been developed to more accurately predict and study the joint and musculotendon forces.[3–5] Some of the errors introduced in the net joint forces and muscle forces associated with inaccurate joint locations of the hip[6,7] and knee[8] have also been described. Challis and Kerwin describe the influence of many of the assumptions on the resultant joint moment for an elbow flexion activity.[9] Though the results were most sensitive to uncertainties in the derivatives

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of the position data, they did note that "the instantaneous joint center or centroid can move throughout the movement, which makes reference to anatomical landmarks very difficult."

There are polycentric exo-prosthetic knees that are designed to have a mobile center of rotation. For the prosthetic user, this mobile center of rotation provides many advantages for ambulation. During stance, the center of rotation is designed to be proximal and posterior to the corresponding anatomical knee location, which increases stability and control. During swing, the center of rotation is then designed to move distal and anterior.

Unlike the human counterparts previously described, the advantage of studying these components is that the underlying mechanics are easier to examine and model. When the mechanics of the joint are unknown, calculation of the center of rotation may be difficult mathematically, as described by Halvorsen *et al*. and Soudan.[10,11] As described mathematically by Greene, Gard *et al*. calculated the instantaneous center of rotation for the four-bar polycentric knee used in this study as the intersection of the posterior and distal linkage.[12,13] When the knee is fully extended, the center of rotation is 9 cm posterior and 18 cm proximal to the proximal-anterior axis. As the knee flexes (with the thigh segment as the frame of reference), the center of rotation arcs forward and distal. At 15° of knee flexion (measured as the angle between the thigh and shank segments), the instantaneous center of rotation moves to approximately 0.8 cm anterior and 2.5 cm proximal to the proximal-anterior axis. The motion of the linkages during swing phase also causes an apparent shortening of the prosthesis, compared to a single axis knee, which allows for increased floor clearance.[12–15] This results in translational and rotational motion, causing apparent variation in the segment lengths that will not be accounted for in the calculations.

We have developed a two-dimensional mode of a commonly used prosthetic knee to study the effect of the moving center of rotation. The two calculations compare the inverse dynamics results (1) when the four-bar mechanism is included to (2) a model where this movement is assumed to be a single point of rotation. Our specific goals are to (i) compare the magnitudes and patterns of the moments at the knee and hip during stance and swing, (ii) compare the magnitudes and patterns of the power at the hip during stance and swing, (iii) describe how other prosthetic componentry may also influence inverse dynamics calculations.

II. METHODS

To demonstrate the effect of a mobile center of rotation on the calculations related to inverse dynamics an experiment was performed on one trans-femoral amputee walking with a polycentric knee. Data were collected to compare the differences in the analysis when ignoring the mechanism of the 4-bar linkages and when including the mechanism of the 4-bar linkage.

For the first case, the mechanism of the four-bar linkages was ignored. A standard inverse dynamics analysis was performed using the anterior proximal knee axis as the knee center of rotation. The two-dimensional equations relating proximal and distal moments and forces for each segment, using D'Alembert's principle, are given by:[2]

$$\mathrm{Fp_x} = \mathrm{m} \cdot \mathrm{acm_x} - \mathrm{Fd_x}$$

$$\mathrm{Fp_y} = \mathrm{m} \cdot \mathrm{acm_y} - \mathrm{Fd_y} - \mathrm{m} \cdot \mathrm{g}$$

$$\mathrm{Mp} = \mathrm{I}\alpha - \mathrm{Md} - (\mathbf{R}\mathrm{d} - \mathbf{R}\mathrm{cm}) \times \mathbf{F}\mathrm{d} - (\mathbf{R}\mathrm{p} - \mathbf{R}\mathrm{cm}) \times \mathbf{F}\mathrm{p}$$

F is the reaction force, M is the reaction moment, R is a position vector in the global coordinate system, a is the linear acceleration, alpha is the angular acceleration, the p and d subscripts indicate proximal or distal, cm is the center of mass, x and y are the directional components, m is the mass, and I is the moment of inertia about the center of mass. The negative of these values are then used as input for calculating the proximal forces of the next segment (equal and opposite forces).

For the second case, the polycentric knee kinetics are more complex. By dividing the system into the separate linkages, there are an additional 3 joint forces and moments (4 axes of rotation at the knee instead of 1). Each rigid body (shank, anterior knee linkage, posterior knee linkage, and thigh) produces three equations (M, Fx, and Fy) resulting in 12 equations. However, there are now 15 unknowns: M, Fx, and Fy each for the 2 distal knee joints, proximal knee joints and the hip. In order to solve the equations, assumptions were made about the moments produced by the knee linkages. Since friction for swing is adjusted through the two anterior joints, one assumption made was that the moments at each of the two joints were equal. It was also assumed that the two posterior linkages had moments equal to zero (posterior proximal moment equal to zero and posterior distal moment equal to zero).

The calculations were also performed for a constraint where the moment at the distal and anterior joint equaled zero. Little difference was seen in the curves, which were virtually identical throughout the gait cycle. The curves presented are for the equal moment constraint. Since forces are rarely presented in a clinical gait analysis, those values are not presented.

The subject was a 26 year-old male with a right knee disarticulation level amputation and a left transfemoral level amputation who has been using prosthetic gait as his primary mode of ambulation for over seven years. Approval for this study was granted by the Northwestern University Institutional Review Board and informed consent was obtained prior to data collection.

The total mass of the subject (74.5 kg) was calculated from the average of two static trials standing on one force plate. The subject's height (shod) was measured to be 186 cm. The subject's right prosthesis was not changed and a duplicate of his left prosthesis was used for the study. The left prosthesis was fit with an Otto Bock 3R36 Habermann Modular titanium knee joint with mechanical friction to limit heel rise and an internal extension assist. A SACH foot completed the prosthesis. This knee joint has adjustable linkage positions to customize the stability of the knee joint. For this experiment, the knee was set in the most stable configuration and measurements of the linkage lengths and orientations in the extended position were from Gard.[14] For

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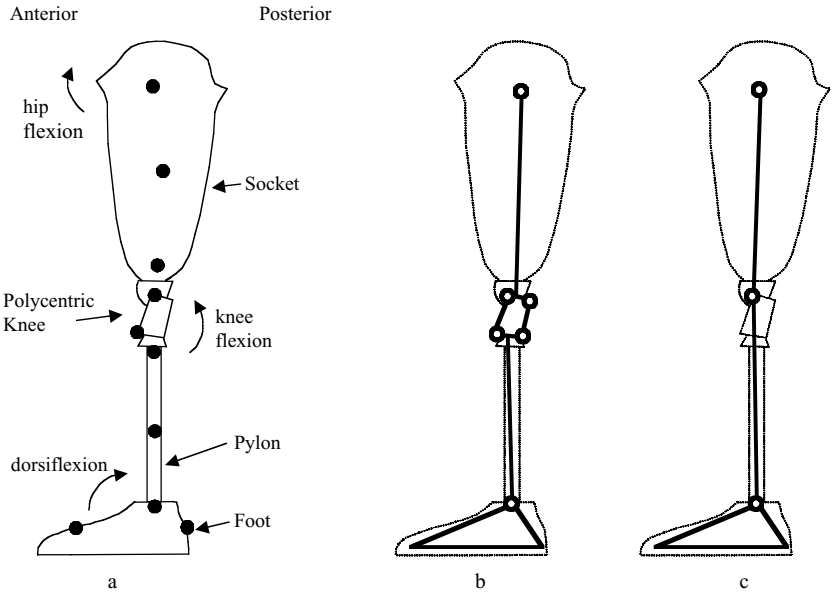


Fig. 1. Sketch of the transfemoral prosthesis showing the marker set and defined linkages for inverse dynamics trials. a) The components of the prosthesis (socket, knee, pylon and foot) are labeled as are various motions described in the text (flexion, etc). Markers are shown as black dots; additional markers were located on the left and right ASIS and the sacrum of the wearer. The linkages of the knee are shown in the approximate orientation of the actual prosthesis. The top linkage is rigid with respect to the socket and the bottom linkage is rigid with respect to the pylon. The knee markers were placed directly over the two front axes of the knee. b) The linkages defined for the four-bar case are given in bold with the joint axes shown as open circles. c) The linkages defined for the single-axis case are given in bold with the joint axes shown as open circles. The proximal-anterior joint axis was assumed to be the knee joint.

the analysis ignoring the four-bar mechanics, the knee axis was assumed to be the anterior proximal joint of the 4-bar mechanism. This joint was chosen because it is the joint recommended by the manufacturer for use in prosthetic alignment. Bench alignment for this knee when using a SACH foot is performed with the weight line (starting point at the bisection of the socket in the sagittal plane at the ischial tuberosity level) falling through this anterior axis and 10 to 15 mm posterior to the midpoint of the foot. An NCOPE (National Commission on Orthotic and Prosthetic Education, USA) registered Prosthetic Resident (the first-named author) dynamically aligned the prosthesis in all planes, including rotation of the knee axis and coronal alignment. Final sagittal alignment was a weight line through the knee and approximately 25 mm posterior to the midpoint of the foot.

Data were collected at 120 Hz using an eight camera system (Motion Analysis Inc.) with infrared reflective markers and AMTI (Advanced Mechanical Technology Incorporated) force plates. Marker data were filtered at 6 Hz using a bi-directional Butterworth filter and derivatives calculated using the central difference method and the filtered data. A data collection trial was considered successful when the left prosthetic foot hit the first force plate and the right foot hit the second force plate. Five successful trials were recorded with the subject walking at self-selected walking velocity. The marker set for the prosthesis is shown in Figure 1 with additional markers placed on the left and right anterior superior iliac spines (ASIS) and the sacrum, in order to define the hip joint. Data were processed using Matlab with algorithms for two-dimensional analysis. Algorithms were derived from those provided by Ton van den Bogert through the International Society of Biomechanics software source. The programs were modified to account for changes in data format, sampling rate, and filtering.

Although the processing for comparison of the knee was 2-D analysis, the hip joint was calculated using 3-D kinematics and the equations given for the Motion Analysis Inc. software. For the calculation of the hip joint center, the pelvic coordinate system is defined with an origin at a point equidistant between the right and left ASIS markers. The y-axis is directed from the left ASIS marker to the right ASIS marker. The x-axis is defined from the sacral marker to the origin and the z-axis is the cross-product of the x- and y-axes. The joint center is then defined as a percentage of the ASIS breadth along each axis: 22% of the distance in the x-direction (posterior), 32% in the y-direction (lateral), and 34% in the z-direction (inferior).[16] The position of the hip joint was calculated in 3-D space and then the 2-D projection was used for the subsequent analysis.

The masses used for the calculations were measured from the prosthesis. The foot was disconnected and weighed. The shank mass included the knee mass for the first case and was divided into two equal segments for the second case. The masses and segment lengths used are given in Table I. The mass for the thigh was calculated from circumferences and lengths of the residual limb. These values were used to calculate a center of mass for the residual limb, which was combined with the mass of the socket.

III. RESULTS

Notable differences were found between the cases. Figure 2 shows the moment calculations. Moments are scaled to the subject's body weight. For completeness, ankle moments are also given. The dotted lines are those for the standard

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Table I. Mass parameters for the prosthetic limb used in the inverse dynamics analyses.

Subject Mass = 74.5 kg	Single-axis calculation	Polycentric calculation
Foot mass (kg)	1.29	1.29
Foot C.O.M. location*	0.5	0.5
Shank mass (kg)	1.0	0.54 shank 0.23 each linkage
Shank C.O.M. location*	0.270	0.5 shank 0.5 each linkage
Thigh mass (kg)	7.13	7.13
Thigh C.O.M. location*	0.62	0.62

*Measured from proximal joint as percentage of segment length.

analysis with the single-axis knee assumption. The solid lines are results when the kinematics of the knee are included. Small differences are seen at the knee early in stance. Since the polycentric knee effectively moves the center of rotation posterior, an increased flexion moment (the extension stop) is expected. During swing, as the joint flexes, more differences are seen. The swing flexion resistance adjustment of prosthetic knee, adjusted to limit heel rise, produces an extension moment. Additionally, the Otto Bock knee has a strong spring-extension assist to ensure that the knee extends fully during the swing phase. Thus the extension moment, seen for most of the swing phase with the polycentric knee case, is expected. However, it should be noted that for the single-axis analysis, the moment quickly shifts to a flexion moment. Toward the end of swing, the knee hits the extension stop at the end of the gait cycle resulting in a flexion moment for both cases.

As the influence of the distal moment propagates proximally, more differences are noted. The prolonged extension moment at the knee during the swing phase influences the hip. The hip moment is also influenced by the mechanics of the linkage motion associated with flexion of the knee. With the influence of the knee kinematics included in the analysis, the initial flexion moment to generate the flexion of the hip for swing is still seen and is increased in magnitude. During swing a small but prolonged hip flexion moment is seen for the single-axis joint assumption model. However, when the knee joint is properly modeled, there is a large hip extension moment seen for most of the swing phase. Both models calculate a strong hip extension moment at the end of the swing phase to slow the motion of the thigh and prepare the leg for the next step.

Joint powers for the hip joint are also presented (Fig. 3). Joint powers, often reported in gait studies, are equal to the joint moment times the joint angular velocity (e.g. knee flexion moment times the derivative of the knee flexion angle). Positive powers are considered to be power generated (positive moment and positive angular velocity) and negative powers, power absorbed. Because of the unique angular definition of the 4-bar knee and the inability to calculate a meaningful joint angular velocity, knee powers are not presented. The hip power has a pronounced difference with respect to model selection. Both cases define the hip joint in the same way. Therefore, the joint angular velocity used for each calculation was the same. However, because the

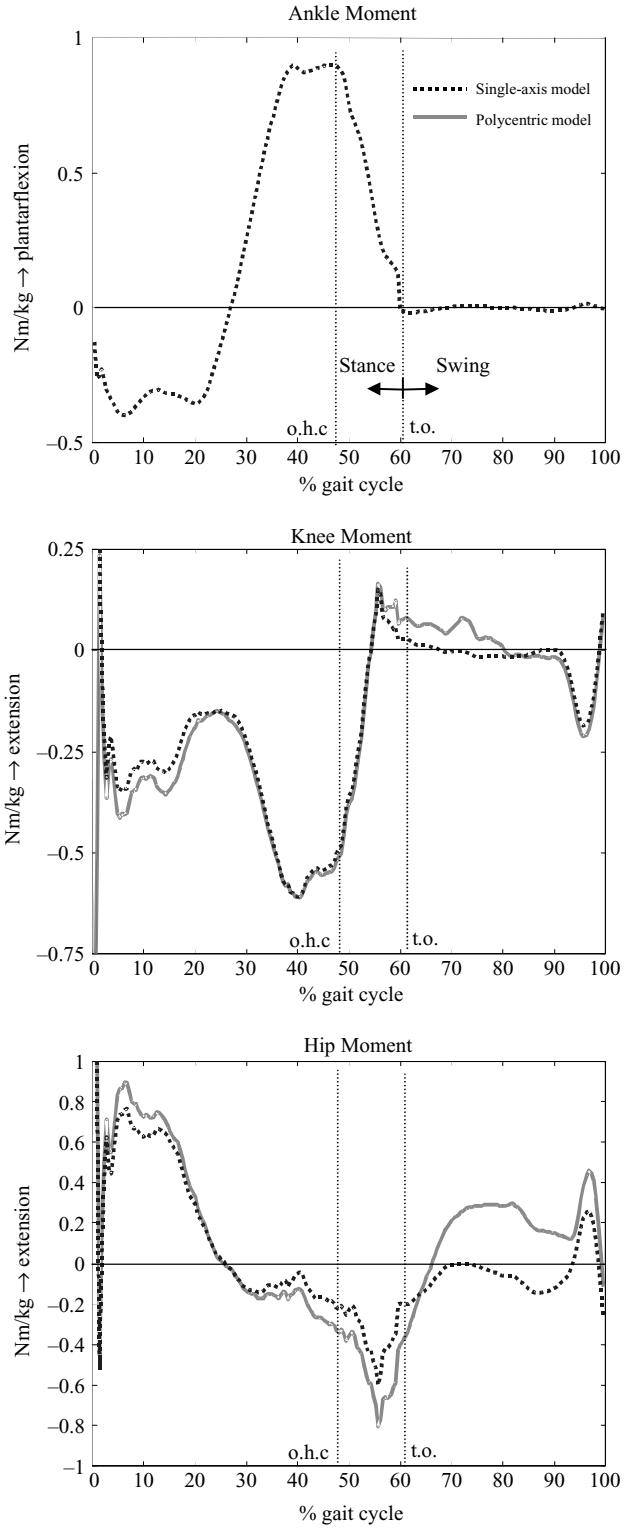


Fig. 2. Moment calculations via inverse dynamics for all three joints (Newton-meters/kilogram). Moments are scaled to the subject body weight and time is scaled to 100% of the Gait Cycle. Toe off (t.o.) occurred at 61% and opposite heel contact (o.h.c.) at 47%. The dotted lines are the results when the joint is assumed to be a single axis of rotation. The solid lines are for the results considering the effects of the polycentric knee mechanism.

hip moment is of a different sign for the polycentric knee condition, the power curve has a shape that is not just a scaled version of the other curve. Instead, because of these different shapes, signs, and magnitudes, interpretation of the values

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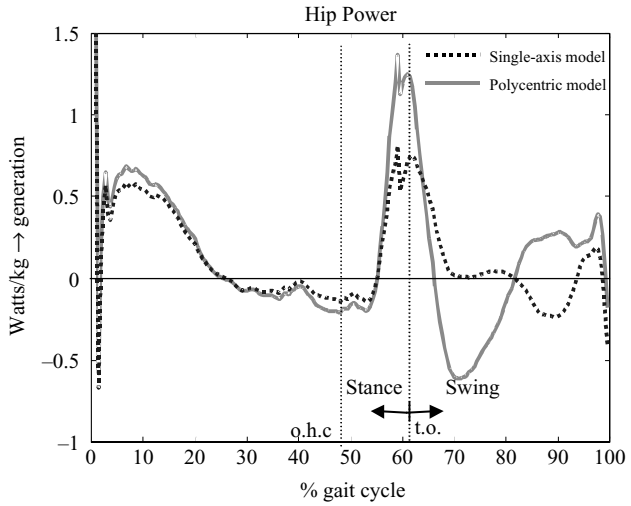


Fig. 3. Power calculations via inverse dynamics for the hip (Watts/kilogram). Powers (joint moment times the joint angular velocity) are scaled to subject's body weight and time is scaled to 100% of the Gait Cycle. Toe off (t.o.) occurred at 61% and opposite heel contact (o.h.c.) at 47%. The dotted lines are the results when the joint is assumed to be a single axis of rotation. The solid lines are for the results considering the effects of the polycentric knee mechanism.

generated by the inaccurate model would result in incorrect conclusions regarding the function of the joint. In order to compare the curves, the values were obtained for each of the four trials at 60, 70, 80, and 90 percent of the gait cycle. These results were then compared using two-sided paired t-tests. Since there are 12 comparisons, the significance level selected was 0.05 divided by 12, or 0.04. The averages, standard deviations, and p-levels are shown in Table II.

Table II. Results for inverse dynamics calculations for each analysis at 60, 70, 80 and 90 percent of the gait cycle. A paired t-test was performed for each variable at each phase of the gait cycle. The mean, standard deviation, and significance level is given. Given the 12 comparisons (3 variables, 4 phases), a two-tailed significance level of 0.004 was chosen (0.05/12). Significant comparisons are presented in bold.

	60%	70%	80%	90%
Knee Moment				
Single-axis assumption	−0.0795	−0.0652	0.0568	0.0268
	±0.0008	±0.0013	±0.0000	±0.0016
Polycentric assumption	−0.0829	−0.0652	0.0024	0.0296
	±0.0000	±0.0013	±0.0000	±0.0003
p-value	.8244	**.0016**	**.0000**	.9039
Hip Moment				
Single−axis assumption	−0.5517	0.0539	−0.1358	−0.2118
	±0.0057	±0.0014	±0.0071	±0.0033
Polycentric assumption	−0.3656	0.2602	0.2862	0.1539
	±0.0161	±0.0017	±0.0031	±0.0041
p-value	0.0455	**0.0003**	**0.0002**	**0.0001**
Hip Power				
Single−axis assumption	1.9725	−0.1396	−0.0362	−0.4202
	±0.1367	±0.0089	±0.0088	±0.0253
Polycentric assumption	1.2617	−0.6919	−0.0101	0.3160
	±0.0723	±0.0076	±0.0299	±0.0332
p-value	.0209	**0.0001**	0.7999	**0.0009**

IV. DISCUSSION

The results of this study show that prosthetic componentry designed to have a mobile center of rotation will affect the results if not accounted for in the analysis. In some cases the prosthetic user and the laboratory staff might not even know the type of prosthetic knee being used without removal of a cosmetic cover. Placement of the markers would then be on top of the cover and would be affected by its movement and would mask the effects of the knee movement. This may be similar to movement of the skin and soft tissue, though with soft tissue it is much more difficult to "remove the cover" and model the underlying mechanics of the joint.[17] Holden *et al*. have shown that the error in knee joint center location differs by less than 11 mm with the use of a percutaneous skeletal tracker compared to surface mounted targets.[18] However, this comparison may not be valid because, while some prosthetic knee joints are single axis joints, many such as the knee under study are not and have a center of rotation that varies more than that of the normal human knee.[11,13]

Some of the assumptions of the knee model were varied to test the results. Mass was assumed distributed between the two segments. Reducing the mass of the knee linkage segments had no discernable effect on the moments. The assumptions of the knee moment distribution were also investigated. Little effect was evident when the moment was assumed to be solely on either of the anterior axes or when it was assumed to be equal between the two anterior axes.

The hip extension moment seen for the polycentric knee case is similar in magnitude to that reported by Winter[2] and used in the swing leg simulation by Piazza and Delp, though slightly prolonged in time. Piazza and Delp showed how different flexion and extension moments at the hip and knee can influence knee flexion in swing.[19] The values reported are between 20 and 30 Nm (scaled to body weight these are equal to approximately 0.26 to 0.40).[2,19] The moments calculated by the Piazza and Delp model were based on muscle models and non-muscular springs (passive tissue) and produced an extensor moment similar in shape and magnitude to the Winter values.[19]

The hip power absorption for the polycentric method is calculated because there is a positive velocity and a negative moment. The single-axis method calculates a positive hip moment. The rapid flexion extension moment, corresponding to the power generated and absorbed at the hip may be explained by the relationship between the moment and power. By rapidly accelerating the hip forward (flexion velocity and flexion moment), there is a positive power. This moment will cause the knee to flex for floor clearance. After the toe leaves the ground there continues to be a flexion velocity (negative); however the moment changes sign. This extension moment acts to slow the thigh and allow a transfer of momentum to the shank, to encourage knee extension. Prior to heel contact, the hip has reached full flexion and begins to extend back. This extension movement, coupled with the extension moment causes a final peak in the power curve. This movement is common in amputees as they pull back to ensure that the knee is fully extended at the next heel contact.

There have been other calculations of inverse dynamics for transfemoral amputee gait. The references with results

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presented in a similar manner (e.g. same foot used and the same calculations presented) can be found in Winter *et al*.[20] They compared the results for an individual using a transfemoral prosthesis to "normals". The results were similar to those presented for the single-axis assumption. Compared to normals they found an increased hip moment (in both extension early in stance and flexion late in stance early in swing), an extension moment at the knee throughout stance and a reduced ankle plantarflexion moment. They found a hip power greater in magnitude than the normals (increase power generation early in stance, increased absorption later in stance and peak in power generation early in swing. The current results were similar to those reported by Winter, though closer to the "normals" in magnitude.

Inverse dynamics models are widely used in robotics and mechanics where they work well if there is complete knowledge of the system being investigated. This method has become very common in clinical gait labs now that measurement systems and computers have become fast and relatively affordable. However, in biomechanics, calculations and interpretations of inverse dynamics data suggest that one must be careful when using this technique. Some clinicians, especially those involved with children with cerebral palsy, rely heavily on moment and power curves, while others may not; as this method gains more widespread use in prosthetics as well as orthotics among clinicians without a technical background, there is a risk the methods will be improperly applied by those without an understanding of the assumptions and possible errors of the model and methods.

A basic assumption of an inverse dynamics analysis is rigid link segments with ideal hinge or ball-and-socket joints. Polycentric knees are specifically designed to have mobile centers of rotation; without specific selection of marker location, these effects may not be discernable. The results of this study may be interpreted to show that prosthetic componentry has little effect on the stance phase of gait. However, the opposite should be true. Due to the low forces experienced during the swing phase, when only the acceleration terms will influence the moment values, small differences would be expected as a result of model variation. Yet significant differences were seen even during double support and swing when the forces terms are very low or zero. The component analyzed in this study remained rigid during the stance phase and only moved during the swing phase. Thus, small differences are seen during stance. However, new prosthetic components are designed to move even during the stance phase. For example, other more complex polycentric knees with six or more linkages have been designed to flex up to 15 degrees during early stance phase. Because of the multiple linkages, the assumption of flexion occurring about a single axis may or may not be valid. Since the knee flexion occurs during a phase of the gait cycle when forces are high, even small errors in the joint axis assumptions may result in significant errors in the moment calculations.

Additionally, shock-absorbing pylons shorten in length when loaded. They may shorten six to twelve millimeters during normal to fast gait, violating the constant length assumption. Also, many prosthetic feet violate rigid segment assumptions (flexible keel prosthetic feet). It is important to remember the limitations and errors that are associated with the inverse dynamics method and that errors are introduced from the assumptions of the model and the measurements.

Further research should investigate the potential magnitude of these other errors; however, components that differ markedly from inverse dynamics assumptions may increase the difficulty of the analysis. Solving the equations of motion using inverse dynamics techniques may require even more thorough knowledge of the mechanism because of the complexity in determining suitable constraint equations to solve the equations of motion.

Therefore, more effort should be dedicated to other methodologies that do not rely on this type of analysis and instead work on the development of models that more accurately describe the system and unique components. For example, forward dynamic models may use the entire musculoskeletal system and muscle control schemes to predict the movement of the body. These models could also be developed with more accurate representations of prosthetic and orthotic components and function. Also, simplified models that predict the net movement of the body rather than the effect of specific joints may aid in understanding the goals of normal human locomotion and how prosthetic and orthotic components may fit into the complete system.

V. CONCLUSIONS

In summary, the results of this experiment show that prosthetic inverse dynamic gait models that do not consider the properties of the components are prone to errors. The largest differences were seen during the swing phase, when the knee was bending. Misrepresentation of the knee joint resulted in differences in the hip joint moment and powers that were not just scaled values of the correct model. The single-axis knee model predicted a hip flexion moment for the majority of swing, but the more accurate polycentric knee model predicted an extension moment that more accurately represented the expectation of the components. These values further influenced the hip power causing curves of a different shape, magnitude, and sign. This consideration is important when performing these experiments and analyses on this population and when interpreting results in the literature. Although for this experiment the errors were seen mostly in the swing phase when the movement of the multiple linkages caused the joint to vary the most from the single-axis assumption, other components that alter the segment length or polycentric knees that flex during stance may cause similar errors during both the stance and the swing phases. The results show that, when comparing different components, care must be taken to accurately model the system so that the comparisons are of the actual performance and are not just errors associated with the models used.

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