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Technical note

# Comparison of methods for the calculation of energy storage and return in a dynamic elastic response prosthesis

Mark D. Geil[a,*], Mohamad Parnianpour[b], Peter Quesada[c], Necip Berme[d], Sheldon Simon[e]

[a]Department of Health and Performance Sciences, Georgia Institute of Technology, Atlanta, GA 30332, USA

[b]Department of Industrial, Welding,and Systems Engineering, The Ohio State University, 210 Baker Systems, 1971 Neil Avenue, Columbus, OH 43210, USA

[c]Department of Mechanical Engineering, J. B. Speed Scientific School, University of Louisville, Louisville, KY 40292, USA

[d]Department of Mechanical Engineering, The Ohio State University, 2075 Robinson Laboratory, 206 West 18th Avenue, Columbus, OH 43210-1107, USA

[e]Department of Orthopaedics, Beth Israel Medical Center, 5th Floor, 170 East End Avenue (at 87th Street), New York, NY 10128, USA

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**Abstract**

The standard method used to calculate the ankle joint power contains deficiencies when applied to dynamic elastic response prosthetic feet. The standard model, using rotational power and inverse dynamics, assumes a fixed joint center and cannot account for energy storage, dissipation, and return. This study compared the standard method with new analysis models. First, assumptions of inverse dynamics were avoided by directly measuring ankle forces and moments. Second, the ankle center of rotation was corrected by including translational power terms. Analysis with below-knee amputees revealed that the conventional method overestimates ankle forces and moments as well as prosthesis energy storage and return. Results for efficiency of energy return were varied. Large differences between models indicate the standard method may have serious inadequacies in the analysis of certain prosthetic feet. This research is the first application of the new models to prosthetic feet, and suggests the need for additional research in gait analysis with energy-storing prostheses. © 2000 Elsevier Science Ltd. All rights reserved.

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## 1. Introduction

The standard joint power method applied to the ankle joint generates a power curve based on the joint angular velocity and the net joint moment (Winter, 1990). This approach utilizes a familiar kinetic measure and enables comparison to normal gait. However, Prince and Winter (1994) have reported the presence of errors in the application of this approach to energy storing prostheses. These prosthetic feet, also known as dynamic elastic response (DER), use a leaf spring keel to store energy in midstance during the gait cycle, returning a portion of that energy at toe off. The standard rigid body analysis model cannot account for the changing center of rotation and assumes no energy loss. Buczek and Kepple (1994) found in normal subjects that the incorporation of segmental endpoint velocities as translational power terms can account for the discrepancy between the instantaneous center of rotation and the fixed joint center assumed by the standard link-segment model. The authors noted that energy from the sum of translational power terms (as measured in normal subjects) is at times greater than the difference in prosthesis energy storage and return reported in the literature.

The purpose of this research was to quantify the potential errors produced by two fundamental assumptions of a conventional analysis of DER feet, namely the assumptions of rigid bodies and fixed axes of rotation. The outcome of this conventional analysis on computed estimates of energy storage and return in a DER prosthesis was contrasted to methods designed to avoid the two fundamental assumptions. These new methods utilized direct measurement of net joint forces and moments and corrections to the axis of rotation.

## 2. Materials and methods

This study was conducted in accordance with the standards of the Ohio State University Human Subjects

* Corresponding author. Tel.: + 1-404-894-9993; fax: + 1-404-894-7593.

*E-mail address:* mark.geil@hps.gatech.edu (M.D. Geil).

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Review Board. Four subjects (Table 1) recruited for the study were active unilateral trans-tibial amputees. All subjects had experience with some type of DER prosthesis. The subjects' residual limbs were fully healed and stable in volume.

Kinematic and kinetic motion analysis was performed on each subject while wearing a custom-ordered Carbon Copy High-Performance (HP) prosthetic foot (Ohio Willow Wood, Mt. Sterling, OH). A six-channel pylon force and moment transducer (PY-6, Bertec, Columbus, OH) was placed at the distal end of the pylon of each subject during testing. Each subject was provided at least a week of accommodation with the HP foot and distal pylon weighting to simulate the weight (383.5 g) of the PY-6 transducer. Prosthesis alignment and fittings were supervised by each subject's respective certified prosthetist.

Reflective markers were placed on 15 body landmarks according to a modified Helen Hayes marker set. Markers were tracked by a five-camera VICON (Oxford Metrics, Baton Rouge, LA) motion analysis system. A 16th marker was placed on the PY-6 transducer. The transducer was connected by cable to a preamplifier, which was connected to a VICON etherbox through Noraxon (Scottsdale, AZ) external data ports. An assistant carried the pre-amplifier behind the subject to limit encumbrance. Ground reaction force was measured with two AMTI (Watertown, MA) force platforms. Kinematic data were collected at 50 Hz while force plate and transducer data were collected at 1000 Hz.

Subjects wore their own normal walking shoes and were instructed to walk at a self-selected comfortable speed. Each subject's starting point was varied to maintain clean footfalls on the force platforms. A trial was accepted when one and only one foot landed completely on one of the platforms. Approximately 10 acceptable walking trials were analyzed per subject.

Two analysis models were compared along with two methods for each, resulting in a comparison of four analysis techniques. The first model (R, indicating only rotational power terms) is the standard analysis of joint rotational power (Elftman, 1939; Winter, 1990), calculated as

$$P_{\text{ankle}} = M_{\text{ankle}}\omega_{\text{ankle}}, \quad (1)$$

where $P$ is the power, $M$ is the joint moment, and $\omega$ is the joint angular velocity. The second model (RT, indicating rotational and translational power terms) incorporates the translational power terms described by Buczek and Kepple (1994) as

$$P_{\text{ankle}} = M_{\text{ankle}}\omega_{\text{ankle}} + F_{\text{ankle}}(v_{\text{d}} - v_{\text{p}}), \quad (2)$$

where $v_{\text{d}}$ and $v_{\text{p}}$ are the velocity of the distal end of the pylon segment and the proximal end of the foot segment, respectively, and $F_{\text{ankle}}$ is the joint reaction force.

Table 1
Subject characteristics[a]

| Subject | Sex | Age (y) | Height (m) | Body mass (kg) | Affected side | Foot length (cm) | Inter-ASIS distance (cm) | Affected side ASIS-trochanter distance (cm) | Affected side leg length (cm) | Socket knee width (cm) | Shoe ankle width (cm) |
|---|---|---|---|---|---|---|---|---|---|---|---|
| 1 | M | 46 | 1.88 | 111.1 | L | 27 | 24.1 | 15.0 | 103.5 | 13.2 | 7.2 |
| 2 | M | 24 | 1.88 | 88.4 | R | 27 | 23.0 | 13.0 | 98.0 | 10.5 | 8.7 |
| 3 | M | 37 | 1.83 | 83.9 | R | 27 | 21.6 | 16.5 | 92.0 | 10.7 | 10.5 |
| 4 | M | 37 | 1.68 | 69.4 | R | 25 | 20.5 | 11.7 | 83.8 | 9.9 | 9.2 |
| Mean | — | 36 | 1.82 | 88.2 | — | 26.5 | 22.3 | 14.1 | 94.3 | 11.1 | 8.9 |
| (Std. dev.) | — | (9) | (0.10) | (17.3) | — | (1.0) | (1.58) | (2.12) | (8.4) | (1.4) | (1.3) |

[a] All subjects were unilateral below-knee traumatic amputees, tested at least 6 months post amputation. All subjects had experience with a DER prosthesis.

For each model, two methods of computation of the force and moment at the ankle were compared. First, joint kinetics were calculated based on inverse dynamics (I). Second, they were measured directly (D) with the pylon transducer. Each of these new techniques, RD, RTI, and RTD was compared to RI, the current standard of analysis.

Position data were filtered using a fourth-order zero-lag Butterworth filter at a cutoff frequency of 7 Hz (Winter, 1990). Ensemble averages were calculated for all trials for each subject, and then normalized by body mass. The root mean square (RMS) was used to quantify the difference between the power curves for each technique. Energy storage/dissipation was calculated as the integral of the midstance negative region of the joint power curve; energy return as the integral of the late stance positive region of the joint power curve.

## 3. Results

Joint power results agreed with Buczek and Kepple (1994) in that the most dominant power term was the dorsi/plantarflexion (sagittal plane) rotational power ($P_x$), while the next most dominant term was the vertical translational power. Results presented were based on the rotational $P_x$ term. Non-sagittal results are detailed in Geil and Parnianpour (1999). Because the dominant vertical component of translational power corresponds spatially to the rotational $P_x$ term (as the large vertical force is the primary contributor to the dorsi/plantarflexion moment), comparisons include only the $Z$ component of translational power.

In general, the inverse dynamics method was found to overestimate the dominant ankle force and moment terms, $F_z$ and $M_x$ (Fig. 1), with larger differences apparent in late stance phase.

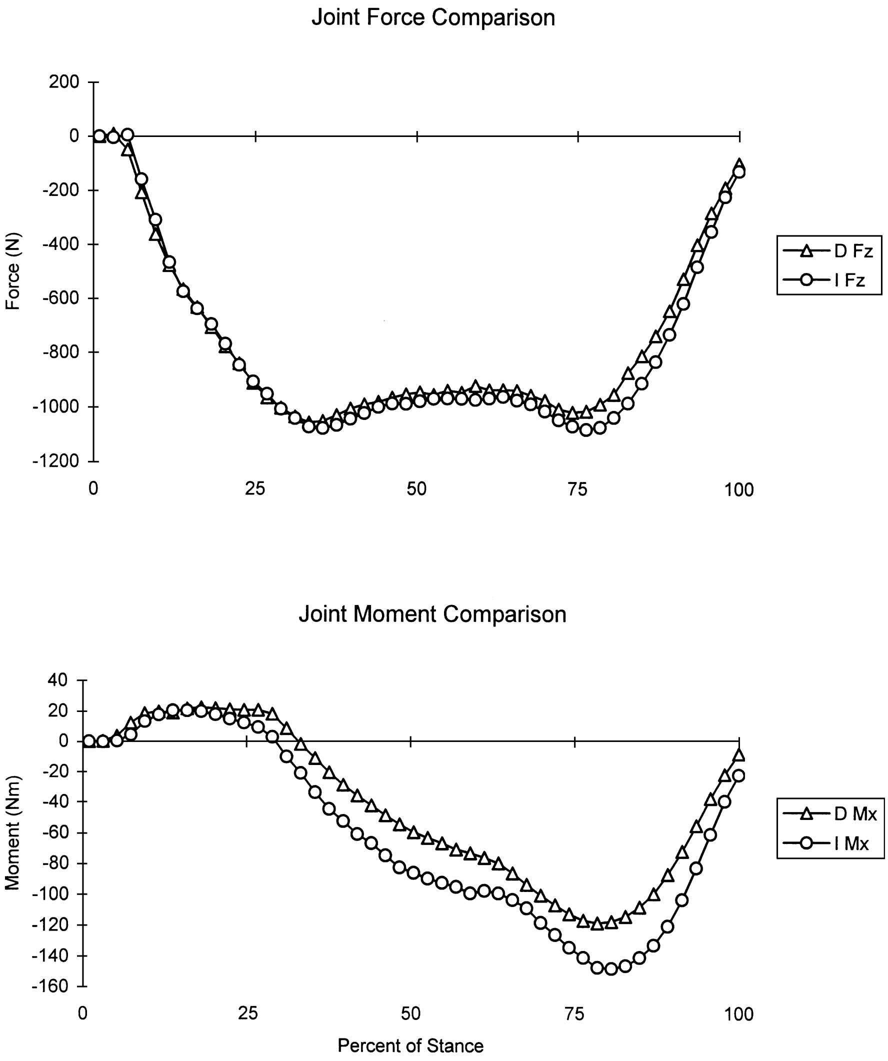


Fig. 1. Representative ankle joint forces (N) and moments (N m) (subject 1, trial 6) measured directly (D) with the PY-6 transducer and through inverse dynamics (I), vs. percent of stance phase.

The impact of model difference (R vs. RT) on normalized ankle power (Fig. 2) is most apparent in early stance, when the foam heel is absorbing energy. Later in stance phase, the curves are more closely paired according to measurement method (I vs. D), with I methods yielding larger absolute values of power. Method RTD revealed the largest RMS difference from RI (ensemble average 0.265 W/kg), while RD showed the smallest difference (0.180 W/kg).

Intrasubject comparisons of prosthesis energy absorption and return (Table 2) between methods reveal the same trends, with I always larger than D. Differences in model (R vs. RT) are generally smaller, and the strong trend in differences is not as apparent. Efficiency differences are varied. Results for subject 4 were unsuitable for energy absorption and return calculations due to large gaps in heel marker data at the critical late-stance portion of the cycle.

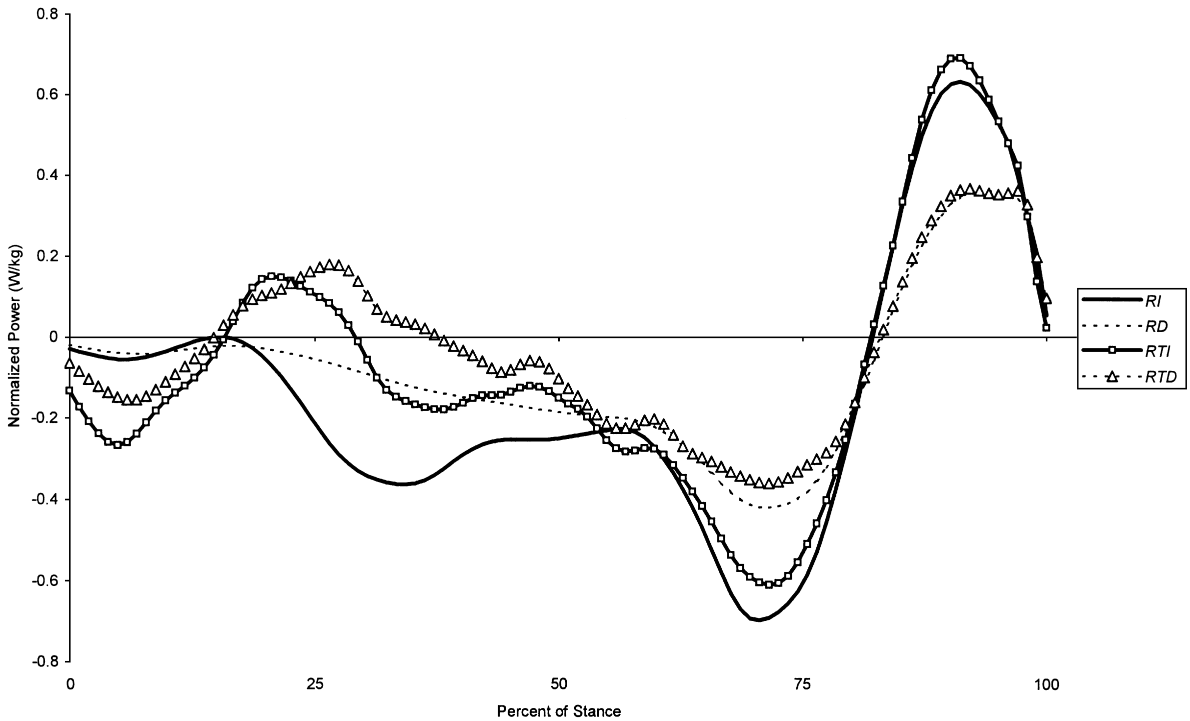


Fig. 2. Normalized sagittal ankle joint power (W/kg) vs. percent of stance phase, ensemble average of all subjects, comparison of each analysis technique: R = rotational power only, RT = rotational plus translational power, D = directly measured ankle kinetics, I = inverse dynamics-calculated ankle kinetics. Comparisons are made of each new method vs. RI, the standard method.

Table 2
Average normalized prosthesis energy absorption, return, and efficiency for subjects 1—3[a]

| Subject | Model | Method | Energy absorption (J/kg) | Energy return (J/kg) | Efficiency (%) |
|---|---|---|---|---|---|
| 1 | R | I | 0.21 | 0.068 | 33.9 |
| | | D | 0.18 | 0.049 | 27.2 |
| | RT | I | 0.14 | 0.082 | 61.6 |
| | | D | 0.12 | 0.058 | 49.4 |
| 2 | R | I | 0.11 | 0.049 | 43.1 |
| | | D | 0.08 | 0.017 | 21.5 |
| | RT | I | 0.12 | 0.044 | 32.4 |
| | | D | 0.11 | 0.017 | 15.6 |
| 3 | R | I | 0.13 | 0.060 | 47.2 |
| | | D | 0.09 | 0.041 | 55.5 |
| | RT | I | 0.13 | 0.060 | 49.5 |
| | | D | 0.08 | 0.042 | 66.0 |

[a]See text for model and method descriptors. Consistent differences were present between model and method in absorption and return; varied differences were found in efficiency.

## 4. Discussion

Comparisons were made in an attempt to theoretically improve calculation of energy storage and return in DER prostheses. Two methods were used to assess the errors in the conventional inverse dynamics method; the methods addressed the joint center of rotation and the foot storage and dissipation of energy.

Results presented are preliminary due to the small sample size and number of feet tested. The foot tested, the Carbon Copy High Performance, uses the fairly standard leaf spring keel, proximal bolt attachment, and cosmetic foam cover found in similar designs from Seattle Limb Systems, Otto Bock, and other Ohio Willow Wood feet. While the subject population did not permit thorough statistical analysis, trends were noted. Our results obtained using the conventional joint power approach appear comparable to those found in the literature; however, amputee results have not typically been normalized (Czerniecki and Gitter, 1991; Gitter and Czerniecki, 1991; Barr and Siegel, 1992; Czerniecki and Gitter, 1992) and direct comparison is difficult. Efficiencies found in this study (Table 2) are typically greater than those previously calculated for SACH feet, but less than those calculated for Seattle and Flex feet.

Directly measured forces and moments were less than comparable joint kinetics determined through inverse dynamics, resulting in similarly reduced energy storage/dissipation and return (Table 2). The inclusion of directly measured forces and moments resulted in less energy stored or dissipated and less energy returned, resulting a smaller efficiency. The source of this discrepancy might be due to either of the theoretical errors in the conventional approach. Both the joint reaction force and the transducer are located at a rigid component of the prosthesis (Geil and Parnianpour, 1997). The center of rotation, which shifts as the leaf spring keel deforms, is actually located much farther distally in the region of deformation. Therefore, errors in moment due to misplacement of the joint reaction force are possible as the moment arm to the joint reaction force locates a force misplaced from the center of rotation. The discrepancy may also be due to viscoelastic elements designed to dissipate (foam heel and cover), store, and return (deflection plates) energy. This energy imbalance is not incorporated in the standard technique (Prince and Winter, 1994). The largest differences in force and moment occur in late stance (Fig. 1), when energy is being stored and dissipated. Power based on direct measurement of force and moment theoretically improves upon this deficiency, but does not entirely eliminate it, as the joint power approach still assumes motion between two rigid segments.

The inclusion of translational power to correct for the joint center of rotation resulted in less energy stored/dissipated, and slightly more energy returned, with a larger efficiency. The dominant vertical translational power term is the second largest power component of the six-degree-of-freedom ankle model, similar to Buczek and Kepple's (1994) findings for normal subjects. However, the impact of the vertical translational power term may be more significant than Buczek et al. found, as the largest power term (dorsi/plantarflexion rotation) is relatively smaller vs. normals. The theoretical interpretation of translational power is less intuitive. The link–segment model used here defines the joint center based on a shared ankle joint marker. This marker is then used to track the distal end of the pylon and the proximal end of the foot during walking. However, the instantaneous center of motion of the foot relative to the pylon may not correspond to the assumed center of motion. The changing center of rotation is reflected as a difference in translational velocity of the proximal foot segment end and the distal pylon segment end when the entire foot is modeled as a rigid segment (Buczek and Kepple, 1994). As a means to account for a changing center of rotation, the addition of the translational power term is particularly useful for comparison of different DER feet, as certain designs exacerbate the change in center of rotation and the error in center of rotation location vs. the anatomical ankle joint.

The enhancement of the foot analysis model and the inclusion of prosthesis instrumentation avoid problematic assumptions associated with analysis of DER prosthetic feet. Directly measured proximal kinetics revealed overestimation of prosthesis energy storage and return in standard inverse dynamics methods. Correction for misplacement of the center of rotation revealed underestimation of energy return efficiency by standard methods. These differences indicate the presence of substantial errors in the application of standard analysis models to DER prosthetic feet.

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