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Heel-region properties of prosthetic feet and shoes

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Abstract—The properties of the prosthetic components prescribed to amputees have the potential to ameliorate or exacerbate their comfort, mobility, and health. To measure the difference in heel-region structural properties of currently available prosthetic feet and shoes, we simulated the period of initial heel-ground contact with a pendulum apparatus. The energy dissipation capacity of the various prosthetic feet ranged from 33.6% to 52.6% of the input energy. Donning a shoe had a large effect. Energy dissipation of a Seattle Lightfoot 2 prosthetic foot was 45.3%, while addition of a walking, running, and orthopedic shoe increased energy dissipation to 63.0%, 73.0%, and 82.4%, respectively. The force versus deformation response to impact was modeled as a hardening spring in parallel with a position-dependent damping element. A nonlinear least-squares curve fit produced model coefficients useful for predicting the heel-region impact response of both prosthetic feet and shoes.

Key words: amputation, artificial limbs, biomechanics, prosthetics, rehabilitation.

INTRODUCTION

Problems with the skin and soft tissue of the residual limb are common reasons why some lower-limb amputees are unable to pursue desired vocational and recreational interests. The repetitive impact loading from heel-ground contact during walking can sometimes lead to residual-limb tissue breakdown and pain. While the intact body has natural mechanisms such as the heel pad and joint movement to attenuate impact forces, the reduced capacity of the amputee forces reliance on prosthetic components for energy dissipation.

Results from in vivo studies [1–5] of heel pad properties have produced significantly different results than ex vivo studies on isolated heel pads [6–8]. The difference is thought to be due to limb and whole body dynamics influencing the response and masking the accurate assessment of heel pad properties [7–9]. Likewise, in vivo comparisons of energy-dissipating prosthetic components may demonstrate similar confounding interactions associated with residual limb and whole body dynamics. Additional experimental variance may be introduced by protocols without an adequate accommodation period necessary to allow amputee adaptation to a new prosthesis.

Biomechanics studies have also examined energy absorption properties of isolated prosthetic feet with and without shoes [10–13], but the studies measuring the properties of the heel region applied velocities much slower than physiologically warranted. For viscoelastic materials, applied velocity is an important independent variable.

Abbreviation: VAPSHCS = Department of Veterans Affairs Puget Sound Health Care System.

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This paper presents results of pendulum impacts intended to simulate the 50 to 100 ms period following initial heel-ground contact of the prosthetic foot during amputee walking. This method of loading the heel region of a prosthetic foot eliminates problems associated with whole body dynamics and human subject variability, but relies on appropriate selection of input velocity and pendular mass to provide sufficient kinetic energy. The response to impact is presented as a means to discriminate differences between prosthetic feet and the effects of shoes. We used a nonlinear viscoelastic model, consisting of a hardening spring in parallel with a position-dependent damper, to provide a theoretical basis for understanding the effects of structural properties and a means to predict the impact response across a range of walking conditions.

METHODS

To measure and model the heel region properties of prosthetic feet and shoes in response to impact, we constructed a pendulum to mechanically simulate the conditions immediately following initial heel-ground contact during walking (**Figure 1**). A pendular mass of 6.6 kg was used to duplicate the effective mass of the stance limb at the instant of heel-ground contact. This mass is less than the 11.6 kg mass used by Aerts et al. [8], but the smaller mass was used to represent the lighter prosthetic limb in comparison to an intact limb. The contact surface of the pendular mass was 12 cm × 12 cm to ensure full heel surface contact at impact. The pendulum was instrumented with an accelerometer (Entran, Fairfield, NJ) to measure the accelerations during and immediately following impact. The acceleration data were double-integrated to obtain position during pendulum contact with the foot. The velocity immediately prior to impact, required for the second integration, was calculated with the use of two fiber-optic photoelectric sensors (Aromat, New Providence, NJ) located 1 cm apart at the base of the pendulum. The optical sensor signals, each conditioned with a Schmitt trigger and sampled at 20 kHz, provided a time difference that allowed calculation of the velocity at impact (providing ±0.01 m/s resolution).

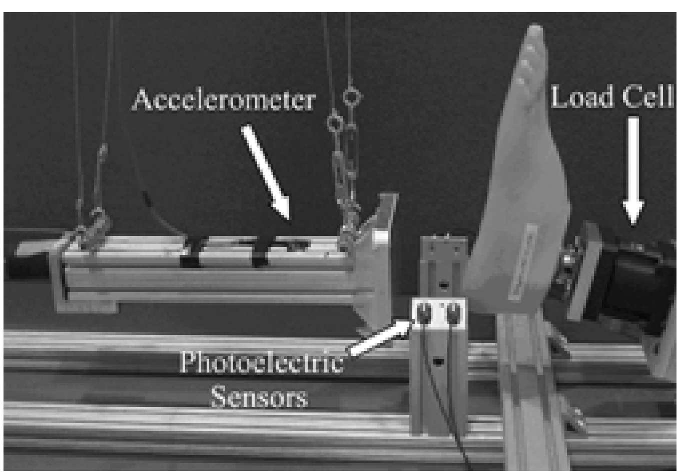

Figure 1.
Pendular mass of 6.6 kg instrumented with accelerometer was used to apply impact loads to each prosthetic foot (and shoe) mounted on load cell.

Seven different prosthetic feet (SACH, Dynamic Plus, SAFE II, Seattle Lightfoot 2, Vari-Flex, Single Axis, and LuXon Max DP) were tested individually, and one prosthetic foot (Seattle Lightfoot 2) was tested with three different shoes (**Table 1**). The prosthetic feet were chosen based on current prescription practice at the Department of Veterans Affairs Puget Sound Health Care System (VAPSHCS) and for the purpose of comparison with other biomechanics studies in the literature. The walking and running shoes were selected as inexpensive, representative models of shoes worn by VAPSHCS patients. The orthopedic shoe tested is occasionally prescribed to patients with a foot deformity and a high probability of foot ulceration.

Each prosthetic foot was neutrally aligned with a standard four-hole pyramid adapter and then angled upward at 20° to simulate the angle of the shank at initial heel-ground contact. This assembly was fastened to a load cell (Advanced Mechanical Technology, Inc., Watertown, MA) on reinforced concrete at the base of the pendulum. The load cell and accelerometer signals were low-pass filtered at 100 Hz with a two-pole Butterworth filter (Measurements Group, Raleigh, NC) and sampled at a rate of 1,260 Hz. The structural assembly was found to have flat frequency response to 100 Hz with a small resonance peak at 120 Hz. The release point of the pendulum was varied to achieve impact velocities of 0.2 m/s, 0.4 m/s, and 0.6 m/s, simulating the potential range of foot velocities experienced during walking [5,14–16].

The choice of pendular mass and impact velocities provides an impact kinetic energy ranging from 0.13 J to 1.17 J. This range, intended to simulate walking, is somewhat lower than the higher kinetic energy used by Aerts et al. [8] (0.80 J to 6.53 J) and Kinoshita et al. [1] (1.30 J

Table 1.
Study prosthetic feet and shoes.*

Test Variable	Description (Manufacturer/Distributor)
Prosthetic Foot	
SACH	SACH Foot with Toes for Men (Otto Bock, Duderstadt, Germany). Suitable for individual up to 125 kg.
Dynamic Plus	1D25 Dynamic Plus Foot (Otto Bock, Duderstadt, Germany). Suitable for individual up to 100 kg.
SAFE II	SAFE II, adjustable style and standard keel, medium heel density (Forsee Orthopedic Products, Oakdale, CA). Suitable for moderately active individual up to 100 kg.
Seattle	Seattle Lightfoot 2, H7 keel (Seattle Systems, Poulsbo, WA). Suitable for medium active individual from 68 to 91 kg.
Vari-Flex	Vari-Flex®, Category 5, split toe, split heel (Ossur, Reykjavic, Iceland). Suitable for moderately active individual from 78 to 89 kg.
Single Axis	Single Axis Foot, regular deflection bumpers, high toe resistance (Ohio Willow Wood, Mount Sterling, OH). Suitable for individual from 79 to 114 kg.
LuXon Max DP	LuXon™ Max DP (Otto Bock, Duderstadt, Germany). Suitable for K3 ambulator up to 136 kg.
Seattle Lightfoot 2 With	
Walking Shoe	Legacy Double Velcro White, Model P-93375 (E.S. Originals, Inc., New York, NY).
Running Shoe	Reebok Catalon Running Shoe for Men (Reebok International Ltd., Canton, MA).
Orthopedic Shoe	Extra Depth®, Bud Special, Black Hillside (P.W. Minor & Sons, Inc., Batavia, NY).

*All were left foot, size 27, with 3/8 in. heel.

and 2.16 J), who intended to simulate running. Interestingly, when Kinoshita et al. attempted a higher kinetic energy of 3.24 J, their subjects complained of pain, indicating a potential upper boundary for the experimental conditions.

Measures of interest to compare the different prosthetic feet and shoes include magnitude of the peak force, peak deformation, and energy dissipation. Energy dissipation (D_s) is defined as the ratio of dissipated energy per loading-unloading cycle to input energy:

$$D_s = \left(\frac{\oint F dx}{\frac{1}{2} m v^2} \right) \times 100 \ , \quad (1)$$

where F is the force (N) in response to impact, x is the deformation (m), m is the pendular mass (kg), and v is the velocity at impact (m/s).

To provide insight into the effects of differing structural properties between prosthetic feet, we modeled the prosthetic foot as a nonlinear spring in parallel with a position-dependent damper:

$$F = a x^b + \mathrm{sign}(\dot{x}) c x^d |\dot{x}|^e \ , \quad (2)$$

where a and exponent b are properties of a hardening spring, c and exponents d and e are properties of a position-dependent damper, and $\dot{x}$ represents the rate of deformation (m/s). The sign ($\dot{x}$) term is defined as 1 when $\dot{x} > 0$, 0 when $\dot{x} = 0$, and –1 when $\dot{x} < 0$. Varying the spring coefficients (a and b) alters the elastic energy storage, while varying the damper coefficients (c, d, and e) alters the energy dissipation (**Figure 2**). We used a nonlinear least-squares curve fit algorithm (MATLAB, Mathworks, Natick, MA) to determine model coefficients from experimental data with a pendulum impact velocity of 0.4 m/s. The algorithm uses initial estimates (i.e., guesses) for model coefficients and iterates to minimize the least-squares error between the experimental data and the model prediction. The solutions were found to be robust to variation of the initial estimates. The capability of the model to predict energy dissipation in response to impact was compared to experimental results at all three impact velocities.

This model was chosen based on preliminary observations of the response to impact. In general, the preliminary force-deformation curves revealed a hysteretic loop whose mean value was found to increase with deformation at a rate greater than justified by a direct proportion. Additionally, the hysteretic loop was single valued at zero deformation and at zero velocity (peak deformation), indicating a position- and velocity-dependent damping element [17].

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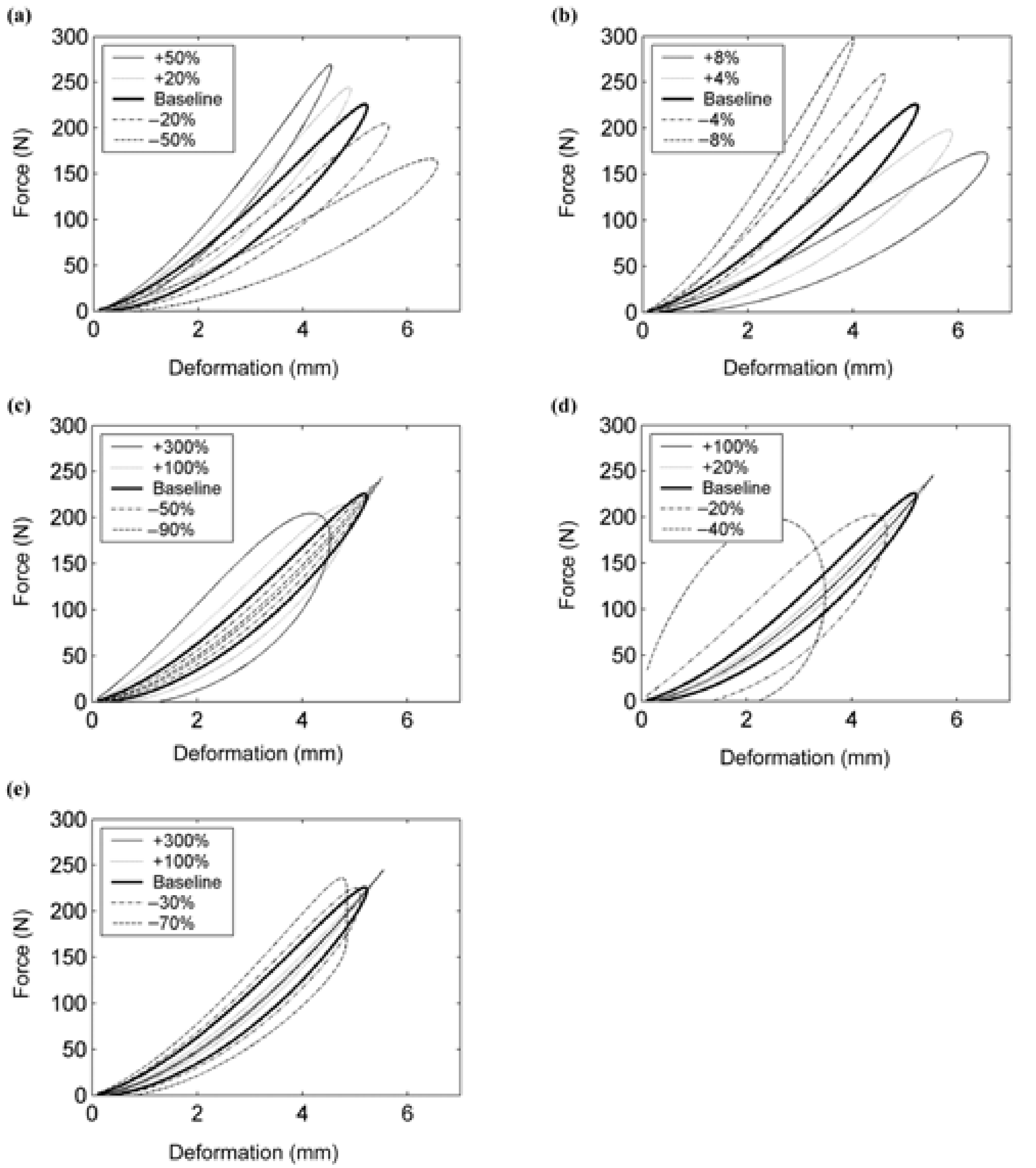

Figure 2.
Effect of varying model parameters on force versus deformation response to impact. Baseline model coefficients were $a = 1 \times 10^6$, $b = 1.60$, $c = 2 \times 10^4$, $d = 1.00$, and $e = 1.00$. Effects of varying model coefficients a, b, c, d, and e are shown in **(a)**, **(b)**, **(c)**, **(d)**, and **(e)**, respectively. Percentage variations were arbitrarily chosen to reveal sensitivity.

Exploration of an exponent for the velocity term showed smaller differences between experimentally measured energy dissipation and model predictions for prosthetic foot with shoe conditions but not for the prosthetic foot alone. Inclusion of this coefficient is necessary to explain the larger hysteretic loops exhibited by the shod prosthetic foot in response to impact. The velocity term exponent was constrained to unity for curve fits of the prosthetic foot alone.

RESULTS

The impact response at 0.4 m/s (**Figure 3**) revealed the SACH foot to have the largest peak force, followed in order by the Dynamic Plus, SAFE II, Seattle, Vari-Flex, Single Axis, and the LuXon Max DP (**Figure 3(a)** and **(c)**). In general, large peak forces were coupled with small deformations across all three tested velocities. The peak force of the SACH foot was nearly twice as great as the LuXon Max DP, while its peak deformation was somewhat less than half. As impact velocity increased (or decreased), the peak force and deformation also increased (or decreased) as expected (**Table 2**). However, the heel-region properties of some of the feet resulted in a reordering of the peak force rank. That is, at the lowest impact velocity (0.2 m/s), the Dynamic Plus (99 N) had a higher peak force than the SAFE II (89 N), and the Seattle (86 N) was higher than the Vari-Flex (77 N). At the highest impact velocity (0.6 m/s), the SAFE II (359 N) had a slightly higher peak force than the Dynamic Plus (357 N), while the Vari-Flex (345 N) exhibited a greater peak force than the Seattle (340 N). Both the Vari-Flex and LuXon Max DP feet exhibited small but difficult to quantify resonance effects, observed as a slight oscillation in the loading and unloading branches of the hysteretic loop (**Figure 3(a)**).

All the feet exhibited clockwise hysteretic loops indicative of energy dissipation. Because of the nature of the materials and the geometry of the prosthetic feet, the percentage of energy dissipation did not remain constant with an increase in impact velocity (input kinetic energy). Some feet had increased energy dissipation (SACH, SAFE II, and Single Axis) with increased impact velocity, while the LuXon Max DP energy dissipation decreased (**Table 2**). The lowest energy dissipation was the 0.2 m/s impact on the SAFE II foot (33.6%), and the highest was for the Single Axis at 0.6 m/s (52.6%).

Placing a shoe on the Seattle foot had a large effect on the impact response (**Figure 3(e)** and **Table 2**). Wearing either a running or an orthopedic shoe increased the peak force in comparison to the prosthetic foot alone for all three impact velocities. The walking shoe increased the peak force only at 0.2 m/s. The peak deformation decreased for all shoes at each velocity, except for the walking shoe at 0.6 m/s. All three shoes resulted in greater energy dissipation at all impact velocities than without a shoe (**Figure 3(e)** and **Table 2**). For example, while the Seattle foot alone absorbed 45.3 percent of the input energy at 0.4 m/s, energy dissipation was increased to 63.0 percent in conjunction with the walking shoe, 73.0 percent with the running shoe, and 82.4 percent with the orthopedic shoe.

Across the range of forces and deformations expected to occur during the first 50 ms to 100 ms of heel-ground contact, the nonlinear elastic element of the model was shown (**Figure 2(a)** and **(b)**) to be very sensitive to changes to the position-dependent exponent coefficient (*b*) and somewhat less sensitive to changes to the proportional coefficient (*a*). Increasing the proportional coefficient (*a*) or decreasing the position-dependent exponent coefficient (*b*) results in higher peak forces for the same kinetic energy input. The position-dependent damping element (**Figure 2(c)**, **(d)**, and **(e)**) was shown to be most sensitive to changes to the position-dependent exponent coefficients (*d*) and relatively insensitive to changes to proportional (*c*) and velocity-dependent exponent coefficients (*e*). Decreasing the proportional coefficient (*c*) or increasing the position-dependent exponent coefficient (*d*) results in higher peak forces. The response to changes in the velocity-dependent exponent coefficient (*e*) was more complex. Increases or decreases from the baseline value both resulted in higher peak forces.

For the 0.4 m/s impact velocity (the condition from which the model was derived), the model underpredicted the energy dissipation by a difference of no more than 6 percent for prosthetic feet alone. At 0.2 m/s, the model underpredicted prosthetic foot energy dissipation by a somewhat larger amount, while at 0.6 m/s the model slightly overpredicted energy dissipation. For the prosthetic foot and shoe combination, the model again underpredicted energy dissipation but by a larger amount than the foot alone for each velocity except for the walking shoe at 0.6 m/s. When the model was used to predict forces and deformations at 0.4 and 0.6 m/s, it tended to predict somewhat smaller magnitudes for the prosthetic feet alone and the shoe-foot combinations (see **Figure 4** for representative results).

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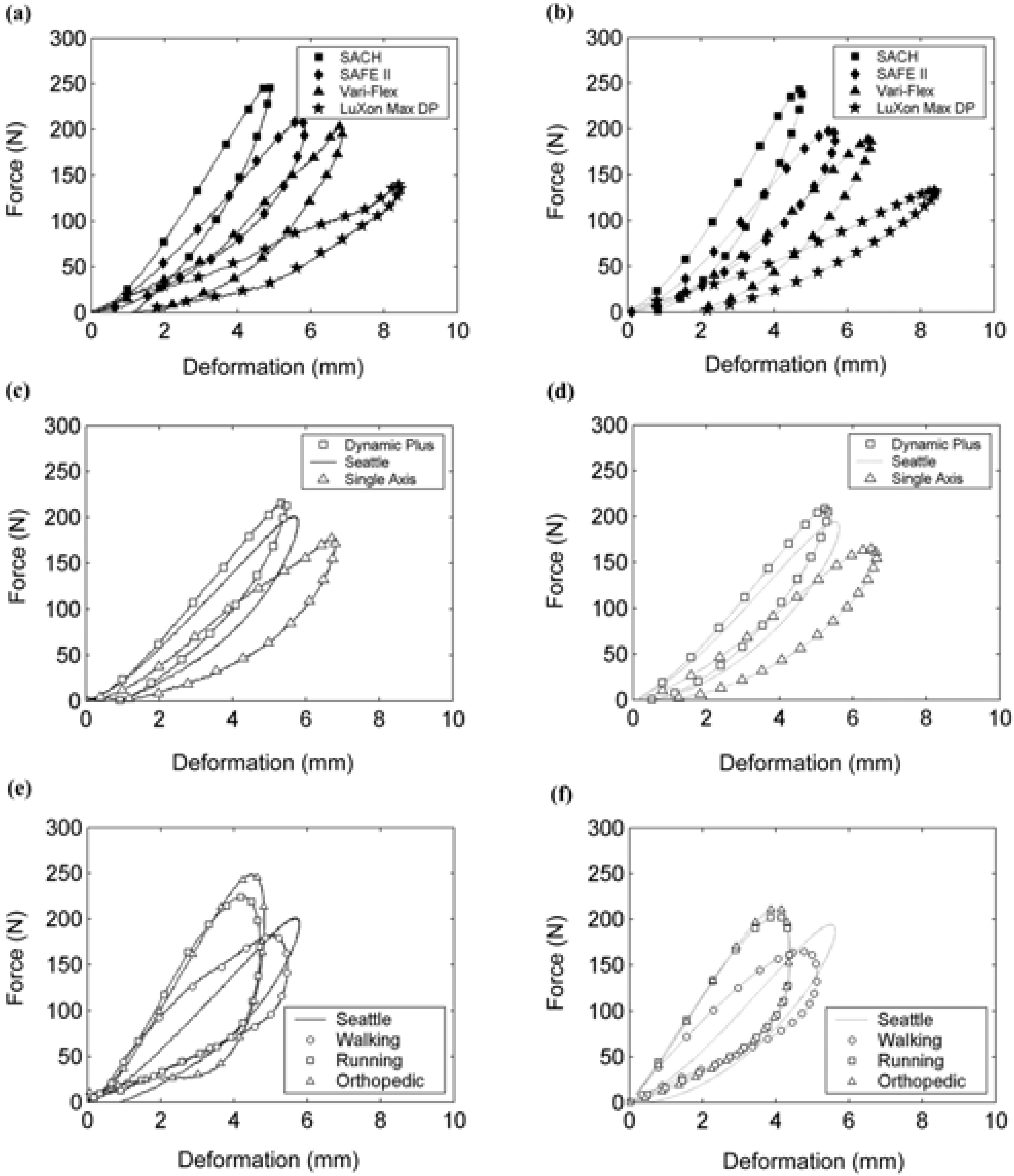

Figure 3.
Experimental force [**(a)**, **(c)**, and **(e)]** versus model force [**(b)**, **(d)**, and **(f)**] versus deformation curves for pendulum impact with initial velocity of 0.4 m/s. Results for SACH, SAFE II, Vari-Flex, and LuXon Max DP feet are shown in **(a)** and **(b);** Dynamic Plus, Seattle, and Single Axis feet in **(c)** and **(d)**; and Seattle foot shod with walking, running, and orthopedic shoe in **(e)** and **(f)**.

Table 2.
Peak force, peak deformation, and energy dissipation as percentage of input energy at each impact velocity for seven different prosthetic feet and three different shoes.

Test Variable	Peak Force (N)			Peak Deformation (mm)			Energy Dissipation (%)		
	0.2*	0.4*	0.6*	0.2*	0.4*	0.6*	0.2*	0.4*	0.6*
Prosthetic Foot									
SACH	111	249	405	2.5	4.9	6.8	34.0	40.1	40.5
Dynamic Plus	99	216	357	2.9	5.5	7.6	35.8	39.8	38.1
SAFE II	89	210	359	3.1	5.8	8.3	33.6	40.4	42.3
Seattle	86	202	340	3.3	5.8	8.0	46.4	45.3	48.4
Vari-Flex	77	205	345	3.7	6.9	9.0	39.8	47.5	44.2
Single Axis	77	178	309	3.7	6.8	9.3	48.7	52.0	52.6
LuXon Max DP	56	141	257	4.4	8.5	11.9	46.3	41.3	36.0
Seattle Lightfoot 2 With									
Walking Shoe	96	183	265	2.7	4.7	8.3	59.0	63.0	60.1
Running Shoe	114	226	342	2.5	5.5	6.6	71.2	73.0	73.9
Orthopedic Shoe	108	250	395	2.6	4.8	7.1	78.1	82.4	86.5

*Velocity (m/s)

DISCUSSION

The properties of the prosthetic components prescribed to lower-limb amputees have the potential to ameliorate or exacerbate their comfort, mobility, and health. The results presented here are intended to aid in prosthetic prescription by providing quantitative properties of prosthetic feet and shoes without the complicating effects of whole-body dynamics or human subject variability. Fitting the experimental data to a nonlinear model provides a means for intuitive understanding and further computational comparative studies. Other researchers have reported in vitro measures of prosthetic heel elastic properties [10,11,18,19], but all used quasistatic methods, in contrast to the dynamic method used here, to simulate the period immediately following initial heel-ground contact during walking.

A limitation of the ballistic approach used here is that both the effective mass (6.6 kg) and the angle of the prosthetic pylon (shank angle, 20°) were held constant during each experiment. Both of these values vary throughout the gait cycle during amputee locomotion. However, because the period where transient forces are frequently observed is within 50 ms to 100 ms following initial heel-ground contact, the structural properties that govern impact response must be measured with an apparatus that applies the appropriate kinetic energy to the heel region using in situ conditions over a short duration. Further, in the study of the human response to impact loads of locomotion, Denoth found a single effective mass could accurately predict the impact peak force of a 61 kg barefoot runner with an effective mass of 8 kg, if the time studied was constrained around the period of impact [20]. A number of investigators have used Denoth's result to develop ballistic, single-mass methods to measure and model human heel properties during running [1,5,8,15], such as the pendular apparatus used here.

Interestingly, the force versus deformation responses of the various prosthetic feet were rather evenly distributed across the range, providing the prosthetist with significant flexibility in selecting the most appropriate foot for a particular patient. The peak impact force exhibited by the SACH foot versus deformation was substantially greater than the LuXon Max DP foot. Clinically, one might hypothesize that a very compliant heel region (e.g., LuXon Max DP) could reduce knee and hip moments in comparison to a less compliant foot (e.g., SACH). At initial contact, the more compliant heel would be expected to have a larger heel deformation and result in a more rapid anterior progression of the center of pressure from the initial, posterior heel contact point. This rapid anterior progression of the center of pressure reduces the foot-ground reaction force moment arm during loading response, yielding the hypothesized reduction in knee and hip moments. Another hypothesis might examine loading response stability, as a foot with a very stiff heel region

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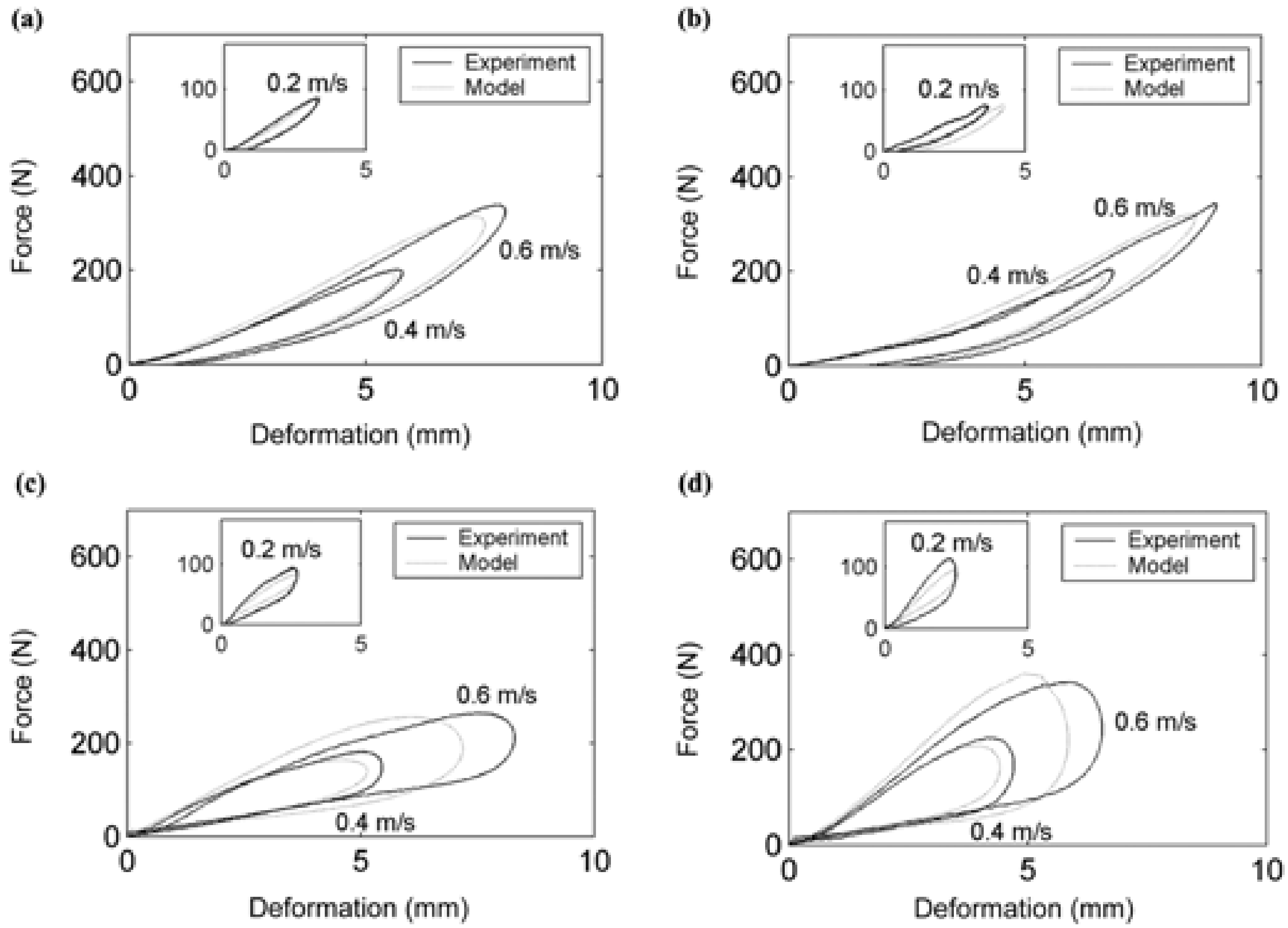

Figure 4.
Experimental force versus model force versus deformation curves for pendulum impact with initial velocities of 0.2 m/s, 0.4 m/s, and 0.6 m/s for **(a)** Seattle, **(b)** Vari-Flex, and **(c)** Seattle shod with walking shoe and **(d)** running shoe.

(e.g., SACH) may prolong the period between heel contact and foot flat and require additional limb stabilizing forces. The prolonged period and need for additional limb forces to maintain a stable loading response may become evident in measures of kinematic symmetry and metabolic costs. With the reported differences in prosthetic feet, our future work will allow testing of these purported benefits, since we can now hypothesize which prosthetic feet can be expected to exhibit a particular response.

The rank order of the feet, with respect to the peak impact force observed, varied with impact velocity. Because the peak impact force occurs just prior to maximum deformation, the damping force can be inferred to be small because the velocity is approaching zero (**Equation (2)**). Thus, the peak force is primarily a function of the nonlinear elastic properties of the heel region. While the force arising from the nonlinear elastic element is not explicitly a function of velocity, greater initial impact velocities (i.e., greater kinetic energy) at impact will generate larger deformations (i.e., greater potential energy) and, hence, greater peak forces. If the heel-region elastic properties were linear, the peak force rank order of the feet would not change as a function of impact velocity. However, because they are nonlinear, the rank order is observed to change.

Across the range of impact velocities expected during walking, energy dissipation by the prosthetic heels ranged from 33.6 percent to 52.6 percent. The LuXon Max DP was the only prosthetic foot whose energy dissipation capacity decreased as impact velocity increased.

This response may have implications for the prosthetic prescription for amputees who may be fit while walking in a clinical environment but participate in vocational or recreational pursuits of a more active (jarring) nature. For patients at risk for residual limb tissue injury, prosthetic feet that dissipate more energy are recommended.

In vitro tests on isolated human heel pads have reported energy dissipation of 32 percent [6], 46.5 percent to 65 percent [8], and 33 percent [7]. The differences between the in vitro human tests may be attributed to different experimental methods and variability associated with a small number of cadaveric specimens (some with vascular deficiencies). These results provide an interesting comparison with the values reported here for the prosthetic heels. The prosthetic heels dissipate approximately the same amount of energy as the biological structure whose function they are intended to replicate.

While there was approximately 18 percent difference between the least amount of energy dissipation (SAFE II at 0.2 m/s) and the greatest (Single Axis at 0.6 m/s), the difference between feet was overshadowed by the effect of shoes. Donning a shoe substantially increased the amount of energy dissipation in comparison to the amount present for a foot without a shoe. These results strongly suggest that patients at risk for residual limb tissue injuries should limit the amount of time spent without shoes, since we hypothesize the increased energy dissipation provided by shoes will reduce the incidence of injury to residual limb soft tissue.

The proposed nonlinear model (**Equation (2)**) consisting of a hardening spring and position-dependent damping element has the capacity to span a wide range of performance. In spite of widely varying peak impact forces and deformations observed in the experimental results, the model provides a reasonable approximation across impact velocities, feet, and shoes. Close examination of model coefficients (**Table 3**) can lead to a more detailed understanding of the observed nonlinear results. For example, the largest proportional elastic coefficient (a), exhibited by the VariFlex foot (more than double its nearest foot [SACH]), might lead one to predict it would have the highest impact force in response to deformation. However, the position-dependent exponent coefficient (b) of the VariFlex is larger than the SACH such that the combined effect is a smaller peak force.

One application of the results presented here is to study residual limb tissue response to impact with a biomechanic model including both biological and prosthetic components. The proposed nonlinear model (**Equation (2)**) of prosthetic foot and shoe heel-region properties could form a subset of such a model. However, a goodness of fit measure, comparing the model to experimental data, is essential for assessment of the potential limitations under varying

Table 3.
Hardening spring and position-dependent damper model coefficients from nonlinear curve fit of experimental data (0.4 m/s condition). Coefficients, when used in **Equation (2)** on page 537, allow prediction of force (N) as functions of position (m) and velocity (m/s). Goodness of fit between model and experimental data is described as difference between energy dissipation predicted by model and energy dissipation observed in experiment as percentage of input energy.

Test Variable	Model Coefficients					Difference Between Model and Experiment Energy Dissipation (% of Input Energy)		
	a	*b*	*c*	*d*	*e*	0.2 m/s	0.4 m/s	0.6 m/s
Prosthetic Foot								
SACH	2,350,000	1.72	20,000	0.91	1.00	–3.4	–2.2	2.2
Dynamic Plus	840,000	1.59	20,000	0.95	1.00	–7.6	–2.8	4.7
SAFE II	520,000	1.53	549,000	1.56	1.00	–13.8	–5.7	3.8
Seattle	710,000	1.59	36,000	1.05	1.00	–16.2	–4.5	–0.5
Vari-Flex	5,380,000	2.05	11,000	0.90	1.00	–2.9	–5.6	0.6
Single Axis	790,000	1.70	50,000	1.14	1.00	–13.1	–4.5	2.4
LuXon Max DP	400,000	1.68	500	0.44	1.00	0.5	–2.3	3.8
Seattle Lightfoot With								
Walking Shoe	16,000	0.90	38,000	1.25	0.73	–24.6	–11.4	2.6
Running Shoe	50,000	1.06	99,000	1.71	0.65	–24.9	–10.2	–1.9
Orthopedic Shoe	85,000	1.15	123,000	1.81	0.61	–27.7	–17.8	–13.7

experimental conditions. As measured by the difference in predicted versus observed energy dissipation as a percentage of input energy (**Table 3**), the impact response of the model was within 6 percent for the 0.4 m/s and 0.6 m/s initial velocity conditions and within 16 percent for the slower 0.2 m/s condition. The largest differences between the model predicted and experimentally observed energy dissipation were for the shod condition at the slowest impact velocity, but importantly, differences decreased with increasing impact velocity. Without impact velocity data from amputee subjects, it is difficult for one to determine if this discrepancy is a significant limitation of the model. A less active sample population (e.g., older amputees) can be expected to walk slower and perhaps exhibit lower impact velocities. In this case, the difference between the model prediction and experimental data could be considered significant (approximately 25%, **Table 3**), and additional model terms would be necessary to improve accuracy. However, it is also plausible that these subjects might have a higher vertical velocity component at impact because their reduced musculature may yield a lesser ability to control prosthetic limb velocities. At higher velocities, the difference between the model prediction and experimental data was much less (footwear dependent, **Table 3**).

Several investigators have used a hardening spring and position-dependent damper to describe the behavior of the shoe-intact foot system [21,22], but no models of the shoe-prosthetic foot system exist in the literature for comparing differences between heel structures. Models of the intact heel pad alone have also been developed. Pain and Challis [9] used a polynomial equation describing a hardening spring with position-dependent viscous damping to model the isolated human heel pad data of Aerts et al. [8]. This polynomial model is somewhat more complicated than the model proposed here, and its only disadvantage is the necessity to solve for additional coefficients. Gefen et al. used a linear elastic spring in parallel with a position-dependent damper to model the in vivo human heel pad [3]. Their results, in contrast with the prosthesis data presented here, did not exhibit a hardening spring.

To describe the forefoot region of prosthetic feet, Geil [12] proposed a linear viscoelastic model and used an iterative technique to estimate model coefficients from a combination of stress-relaxation, creep, and constant strain rate experiments. The slow deformation rates (0.01 m/s maximum) used for calculation of model coefficients yielded an accurate depiction of the forefoot load response to walking for dynamic elastic response feet with a solid ankle, but were not intended for and are likely to be insufficient to model the impact response of the heel region.

One specific aim in developing this model is to use it to understand the relative contribution of the various prosthetic components that comprise a prosthetic limb in attenuating transient forces arising from foot-ground contact. We are interested in discovering which component is the most effective intervention in the system and which component offers the greatest potential for improving effectiveness. Since each element of the prosthesis-limb system acts in series (footwear, prosthetic foot, pylon, socket and liner, and residual limb tissue), a more complex, biomechanic model composed of the biological and prosthetic components (subsystem models) is necessary to predict effects of element properties (e.g., prosthetic foot and footwear) on forces transmitted to the residual limb tissue (i.e., the site of injury) or the predicted foot-ground reaction force. The results presented here suggest that the prosthetic foot and footwear model is accurate at predicting energy dissipation for heel-ground contact velocities ranging from 0.4 m/s to 0.6 m/s, likely relevant for an active amputee while walking at their self-selected speed on level or downhill grades and perhaps stepping off curbs and going down stairs. The model underpredicts energy dissipation at slower heel-ground contact velocities (i.e., 0.2 m/s), conditions that may occur while walking slowly on level or uphill grades and perhaps up curbs and stairs.

CONCLUSION

This paper presents results of pendulum impacts simulating the heel-ground contact of the prosthetic heel during amputee walking. The energy dissipation capacity of the various prosthetic feet ranged from 33.6 percent to 52.6 percent of the input energy. Footwear had a large effect on energy dissipation: the energy dissipation of one prosthetic foot was 45.3 percent, while the addition of walking, running, and orthopedic shoes increased energy dissipation to 63.0 percent, 73.0 percent, and 82.4 percent, respectively. The impact response results (force versus deformation curves) suggest a nonlinear viscoelastic model consisting of a hardening spring in parallel with a position-dependent damper. A nonlinear least-squares curve fit reveals model coefficients predictive of impact response over a range of velocities. Quantifying prosthesis properties through measurement and modeling is an important step in improving the function and performance of prosthetic limbs to meet the needs of amputees.

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