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Dr Herr received support from the Schaffer Foundation and Össur Inc. during the period of performance of the study. Dr Herr is one of many inventors on two U.S. patents that cover the technology utilized by the Össur Rheo prosthetic device. The patents are owned by the Massachusetts Institute of Technology.

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RESEARCH ARTICLE

A Clinical Comparison of Variable-Damping and Mechanically Passive Prosthetic Knee Devices

ABSTRACT

Johansson JL, Sherrill DM, Riley PO, Bonato P, Herr H: A clinical comparison of variable-damping and mechanically passive prosthetic knee devices. Am J Phys Med Rehabil 2005; 84:563–575.

Objective: Although variable-damping knee prostheses offer some improvements over mechanically passive prostheses to transfemoral amputees, there is insufficient evidence that such prostheses provide advantages at self-selected walking speeds. In this investigation, we address this question by comparing two variable-damping knees, the hydraulic-based Otto Bock C-leg and the magnetorheological-based Össur Rheo, with the mechanically passive, hydraulic-based Mauch SNS.

Design: For each prosthesis, metabolic data were collected on eight unilateral amputees walking at self-selected speeds across an indoor track. Furthermore, kinetic, kinematic, and electromyographic data were collected while walking at self-selected speeds across a 10-m walkway in a laboratory.

Results: When using the Rheo, metabolic rate decreases by 5% compared with the Mauch and by 3% compared with the C-leg. Furthermore, for the C-leg and Rheo knee devices, we observe biomechanical advantages over the mechanically passive Mauch. These advantages include an enhanced smoothness of gait, a decrease in hip work production, a lower peak hip flexion moment at terminal stance, and a reduction in peak hip power generation at toe-off.

Conclusion: The results of this study indicate that variable-damping knee prostheses offer advantages over mechanically passive designs for unilateral transfemoral amputees walking at self-selected ambulatory speeds, and the results further suggest that a magnetorheological-based system may have advantages over hydraulic-based designs.

Key Words: Prosthesis, Hydraulic Knee, Magnetorheological Knee, Walking Metabolism

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Recent advances in biomedical engineering have led to the introduction of computer-controlled, variable-damping prosthetic knees for transfemoral amputees. Motivated by the potential of such technology, researchers developed prototype knees,[1–8] and these developments eventually led to marketable devices, such as the Blatchford Endolite Intelligent Prosthesis, the Otto Bock C-leg, and the Össur Rheo. Variable-damping prostheses offer several advantages over mechanically passive designs, including enhanced knee stability and adaptiveness to different ambulatory speeds.[9–11]

Amputees subjectively report that variable-damping prosthetic knees decrease fatigue experienced during ambulation, are easier to maneuver, and allow for smoother movement than mechanically passive knees.[12,13] Following these subjective reports by amputees, several scientific studies have been conducted to quantitatively compare mechanically passive knees with variable-damping devices using measures of gait metabolism, kinetics, and kinematics.

Several studies were performed in the early nineties following the introduction of the Blatchford Endolite Intelligent Prosthesis, a computer-controlled, variable-damping knee device that employs a pneumatic-based strategy for damping modulation. The introduction of this technology was viewed as revolutionary because, contrary to traditional, mechanically passive prosthetic knee designs, the Intelligent Prosthesis allowed for adaptation of knee damping during the swing phase of gait as a function of walking speed. Taylor et al.[14] compared the Intelligent Prosthesis with two mechanically passive prostheses, the Mauch SNS and the Endolite pneumatic swing phase controller. They studied one transfemoral amputee walking on a treadmill at approximately 0.9 m/sec and showed that the Intelligent Prosthesis required about 10% lower oxygen uptake than the passive prosthetic knees. Tests at lower walking speeds were also performed and showed no significant difference in metabolic cost between the mechanically passive and variable-damping knees. Kirker et al.[13] compared the metabolic cost of level ground walking at self-selected speeds when subjects used the Intelligent Prosthesis and a mechanically passive prosthesis. The study involved eighteen transfemoral amputees, and the results indicated that the Intelligent Prosthesis improved gait symmetry over the mechanically passive knee, but no significant difference was observed in metabolic cost at normal, self-selected walking speeds. Buckley et al.[15] performed a study on three unilateral transfemoral amputees to compare the Intelligent Prosthesis with a mechanically passive, pneumatic-based prosthetic knee design (Endolite Stabilized Stance Flex). Compared with the mechanically passive design, the results showed a 5–10% reduction in metabolic cost when individuals walked with the Intelligent Prosthesis at slower and faster walking speeds than their normal, self-selected speed. However, at the normal speed, metabolic rate was not significantly different between the two knee designs.

Similar studies were performed following the introduction of the Otto Bock C-leg in the late nineteen-nineties. The C-leg was considered a step forward compared with the Intelligent Prosthesis because its computer-controlled hydraulic mechanism provided both swing and stance damping control. To assess this new device, Kastner et al.[16] compared the C-leg with two mechanically passive, hydraulic-based designs, the Otto Bock 3R80 and 3R45. The study involved ten transfemoral amputees, and results indicated that the C-leg had smoother kinematics than the passive prostheses. Still further, the authors showed that subjects achieved the fastest time for a 1000-m walk test when using the C-leg. The study concluded that the C-leg provided significant advantages particularly at fast walking speeds. Schmalz et al.[17] tested six transfemoral amputees wearing the C-leg and a mechanically passive prosthesis, the Otto Bock 3C1. Tests were performed by instructing subjects to walk on a treadmill at velocities ranging from 0.5–1.2 m/sec. Results showed a decrease in metabolic cost of about 6.5% for the C-leg over the mechanically passive prosthetic knee. Tests at higher walking speeds were also performed, but the difference in metabolic cost between the two knees was not statistically significant. The fact that there was no significant benefit in metabolic cost when using the C-leg at faster speeds was attributed to the fact that the swing phase damping of the mechanically passive prosthesis was already optimized for walking at those higher speeds.

The studies summarized above suggest that significant advantages are derived when using computer-controlled, variable-damping prostheses compared with nonadaptive, mechanically passive knee devices. For level ground ambulation, the main advantage seems to be the ability of variable-damping knees to adapt to different walking phases and speeds, allowing for early-stance knee flexion and smooth swing phase kinematics. These advantages substantially improve the mobility in individuals who live an active life. Consequently, variable-damping knees are generally prescribed to young and very active individuals. Conversely, the prescription of variable-damping knees is often discouraged in the remaining amputee population because, despite positive subjective reports, it is generally thought that there is insufficient evidence to support the hypothesis that variable-

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damping knees provide advantages at self-selected walking speeds.[18–20] Because it is well established that the metabolic cost of ambulation is significantly higher in lower extremity amputees than in nonamputees, even at self-selected walking speeds,[17,21] researchers have sought improvements in prosthetic knee technology to gain better performance at comfortable walking speeds. This interest for improving prosthetic knee technology has recently generated a novel prosthetic knee device that relies on magnetorheological fluid and a user-adaptive control scheme.[8] This approach has led to a commercially available device called the Össur Rheo knee. Preliminary results in four transfemoral amputees tested at different ambulatory speeds indicated that this variable-damping knee provided users with biologically realistic control of knee flexion during stance and swing.[8]

Based on the enthusiasm generated by these preliminary results on the Össur Rheo knee, we designed a study aimed to investigate the impact of two distinct variable-damping knees, the Otto Bock C-leg and the Össur Rheo, on the metabolic cost and biomechanics of walking at comfortable, self-selected speeds. In addition, we compared these same knees to the Mauch SNS, a mechanically passive hydraulic knee. The Mauch SNS is one of the most common prosthetic knee designs in use today. It utilizes a hydraulic damper that dissipates mechanical energy during joint rotation. Like many commercially available knees, the Mauch SNS passively controls orifice size to adjust how knee damping changes with knee angular velocity.[11] The C-leg is also based on a hydraulic design, but the hydraulic valves are controlled by a microprocessor.[5] In distinction, the Rheo knee utilizes a magnetorheological fluid as the primary torque-producing strategy. Here damping is controlled by varying the magnetic field strength through the modulation of electric current passing through an electromagnet.[8] The C-leg and Rheo have many similarities, but their distinctive torque-producing strategies (hydraulic *vs.* magnetorheological fluid) may yield differences in damping specifically at the minimum, or low-end torque region. Whereas the torque output of the hydraulic-based C-leg has a strong velocity dependence, the Rheo knee employees magnetorhelogical fluid in the shear mode and has a weak torque–velocity dependence due to the shear rate thinning properties of the carrier fluid.[22]

Because of the differences between variable-damping and mechanically passive prostheses, we hypothesize that variable-damping devices offer an improved metabolic economy of gait compared with mechanically passive designs at self-selected walking speeds. Furthermore, we anticipate that the distinct torque-producing strategies of the investigated knees, hydraulic *vs.* magnetorheological, result in differences in metabolic gait economy at self-selected gait speeds. To test these hypotheses, oxygen uptake rate is measured on eight amputees (seven transfemoral and one knee disarticulation) walking at comfortable, self-selected speeds over an indoor track using each of the investigated knee prostheses.

Additionally, we hypothesize that gait biomechanics are significantly different between variable-damping and mechanically passive prostheses and between hydraulic and magnetorheological-based systems. To test this hypothesis, subjects are asked to ambulate at comfortable, self-selected speeds across a level walkway in a motion analysis laboratory. Kinematics and kinetics are estimated using a camera-based system equipped with force platforms for each of the knee prostheses.

Finally, we hypothesize that differences in movement and muscle activation patterns associated with the three knees can be captured by means of wearable sensors.[23] Our interest in this technology originates from the expectation that wearable systems might one day allow for the assessment of prosthetic knees under real-life conditions.[24] To this end, in this investigation we use EMG electrodes and accelerometers to monitor EMG activity and patterns of movement during the laboratory evaluations. EMG sensors and accelerometers could be used as part of a wearable system to monitor amputees in the field.[23] Results are statistically compared across the three knees and associations are sought between the sensor data and kinematics and kinetics derived from the camera-based motion analysis system.

METHODS

Data Collection

Eight unilateral amputees participated in the study. The protocol was approved by the Spaulding Rehabilitation Hospital institutional review board, and written informed consent was obtained from each person before participation.

Amputee participants were experienced at prosthesis ambulation, could ambulate at least at a K3 level (i.e., the patient has the ability or potential for ambulation with variable cadence), and had no other musculoskeletal problems or any known cardiovascular, pulmonary or neurological disorders. The eight participants (seven male, one female) were 29–54 yrs old, 165–194 cm in height, and weighed 61–112 kg. Patient characteristics are summarized in Table 1.

Before the study began, each individual had approximately 10 hrs of acclimatization on each knee prosthesis that was not his or her usual prosthesis. Each amputee subject was asked to commit to three testing sessions. One session was per-

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TABLE 1 Amputee subject characteristics and self-selected walking speed

Subject #	Gender	Age (yrs)	Height (cm)	Weight (kg)	Speed (m/s)	Affected Side	Usual Prosthesis	Socket Type	Cause of Amputation
1	M	53	194	112	0.72	R	C-leg	Suction	Infection
2	M	37	181	96	1.12	R	C-leg	Suction	Trauma
3	M	43	183	87	1.01	R	Rheo	Suction	Infection
4	F	48	165	61	1.23	L	Mauch SNS	Salesian belt with sock fit	Congenital birth defect
5	M	44	178	87	1.16	L	C-leg	Suction	Trauma
6	M	46	190	88	0.99	R	Teh Lin 4-bar knee	Suspension liner with locking device	Cancer
7	M	29	187	91	1.28	L	Endolite	Suction	Cancer
8	M	54	177	77	1.01	R	C-leg	Suction	Trauma

formed using an indoor track to assess metabolic cost of level walking for the three knee systems via oxygen uptake measures. The other two sessions were performed in a motion analysis laboratory to study differences in kinematics and kinetics associated with the three knees. Two prosthetic knees were studied in the first laboratory session and the third knee device was then investigated in the remaining session. The order in which the knee systems were evaluated was randomized. Before any testing, the subjects were fitted with three prosthetic knees by the same prosthetist. Manufacturer recommendations were followed when aligning each knee prosthesis. In addition, each subject used the same prosthetic socket, prosthetic foot, and shoe when testing each knee device. The Össur low profile, high-energy return Allurion foot was used with each knee prosthesis. By using the same prosthetic socket, foot, and shoe for all the tested knees, we assured that differences observed during the study were indeed caused by the different knee prosthetic designs. In other words, we avoided the use of different socket, foot, and shoe systems because they would have played the role of confounding factors. The three knees studied in this investigation are shown in Figure 1. The total mass of

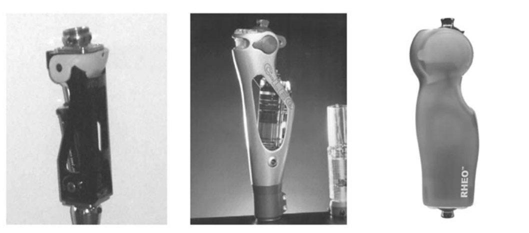

FIGURE 1 *The Mauch SNS, C-leg, and Rheo knee prostheses are shown.*

the prosthetic knee, shank, Allurion foot, and shoe system was 2.71 ± 0.24 kg, 2.72 ± 0.26 kg, and 3.03 ± 0.20 kg (average ± SD) corresponding to the Mauch SNS, C-leg, and Rheo, respectively. Furthermore, the distance between the knee rotational axis and the center of mass of the prosthetic knee, shank, Allurion foot, and shoe system corresponding to the Mauch SNS, C-leg, and Rheo was 27.07 ± 4.59 cm, 27.81 ± 3.53 cm, and 23.94 ± 3.49 cm (average ± SD), respectively.

Before the study began, each individual had approximately 10 hrs of acclimatization on each knee prosthesis that was not their usual prosthesis. In the session performed using the indoor track, oxygen uptake was measured in the amputee participants. Subjects walked at comfortable speeds over a quarter-mile track using a portable, lightweight, breath-by-breath telemetric system (Cosmed K4b2, IT). Before testing, each subject's comfortable walking speed was determined. Then subjects were instructed to walk next to an electrical vehicle programmed to move at their comfortable pace. The same speed was used for all the prosthetic knees and the order of testing was randomized. Time was given to allow each subject to become accustomed to each knee before being tested.

In the sessions performed in the laboratory, kinematic and kinetic data were collected using a motion analysis system (Vicon 512 system; Oxford Metrics, UK). Kinematics were derived by measuring the three-dimensional positions of reflective markers. The markers were placed at the following specific bony landmarks: bilateral anterior superior iliac spines, posterior superior iliac spines, lateral femoral condyles, lateral malleoli, second metatarsal heads, and the calcanei. Additional markers were also rigidly attached to wands and placed over the mid-femur and mid-shank. Kinetics were computed from measures of ground reaction forces derived using two staggered force platforms (AMTI

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Inc., MA) embedded in the walkway. Joint torques and powers were then calculated using a modified version of a standard inverse dynamics model (Vicon Bodybuilder; Oxford Metrics, UK) and were normalized for body weight. This modified version of the model accounted for the altered inertial parameters due to the prostheses, as compared with human anthropometric values found in the literature.[25]

EMG electrodes and accelerometers were also used during the laboratory experiments to monitor muscular activity and patterns of motion, respectively. Active EMG electrodes (Motion Labs, LA) were used to monitor bilaterally the activity of the gluteus maximus and gluteus medius muscles. We chose to monitor these two muscles because their activity has been related to hip control in the sagittal and coronal planes for both normal and amputee gait.[26] In addition to the EMG measurements, uniaxial accelerometers were positioned bilaterally on both thighs and shanks. Accelerometer data were recorded using a Vitaport ambulatory system (Temec B.V., The Netherlands), and these data recordings were synchronized with the Vicon motion capture measurements.

Once the setup was complete, each subject was asked to walk at his/her comfortable walking speed across a 10-m walkway. The amputee subjects were tested in three conditions, each corresponding to a different knee prosthesis. For all subjects, bilateral lower extremity joint kinematic and kinetic data were collected over nine walking trials and averaged for each subject and condition. Each amputee participant was given time to acclimatize to each knee before testing. Amputee subjects were timed to ensure that they walked at the same speed with each prosthesis.

Data Processing and Analysis

Data recorded during indoor track testing were processed to evaluate the impact of different prosthetic knee technologies on the energetic cost of ambulating at a comfortable walking speed. Oxygen consumption ($\dot{V}O_2$) as measured by the breath-by-breath telemetric system (K4b,[2] Cosmed Srl., Italy) was averaged over three minutes of steady-state walking. The rate of oxygen uptake was then calculated by dividing $\dot{V}O_2$ by the mass of each individual.[21] Results were analyzed to test for significant differences among the three knees (repeated measures ANOVA) and pairwise comparisons were performed using the least significant difference test.[27] A significance level of 5% was used for the analysis and estimated *P* values between 5 and 10% were considered indicative of a trend.

Data recorded in the laboratory setting were processed to compare the biomechanical characteristics of ambulation across the prosthetic knee technologies. The following temporal parameters were estimated: walking speed, step time, step length, single support time, and double support time. Kinematic and kinetic trajectories, in the form of joint positions, torques and powers, were characterized by estimating peaks (i.e., maxima and minima) using custom-built software. The peak parameters were selected to capture differences across the prosthetic knees. In addition, work was estimated by integrating hip power to test the hypothesis that a different amount of work is associated with the tested knee technologies. We estimated total positive and negative work contributions for the stance and swing gait periods. Finally, we estimated foot compression on the affected side in the amputee individuals for each of the prosthetic knee technologies. This parameter was derived to capture different dynamic interactions of the prosthesis with the ground. Foot compression was calculated as the difference in vertical height of the shank marker and the malleolus marker. The malleolus marker was positioned on the lateral external surface of the shoe worn over the prosthetic foot. Hence, foot compression as measured here, included both prosthetic heel compression as well as shoe midsole compression. Average values for all the previously defined parameters were estimated for each subject over nine trials and statistically compared via repeated measures ANOVA's and pairwise comparisons performed using the least significant difference test. Estimated *P* values smaller than 5% were considered to be significant, whereas values between 5 and 10% were considered to be indicative of a trend.

EMG and accelerometer measurements were performed during the same laboratory trials utilized to investigate the kinematics and kinetics of gait. EMG activities of the gluteus maximus and gluteus medius muscles were quantified by computing the root mean square value of the data recorded within a gait cycle. Average root mean square values were then estimated for each subject over nine trials and statistically compared via repeated measures ANOVAs and pairwise comparisons performed using the least significant difference test. Significance level and values indicative of a trend were set as explained earlier. Additionally, accelerometer data were processed to explore the hypothesis that differences between prostheses could be measured using an ambulatory system equipped with accelerometer sensors, and to complement the biomechanical data measured with the stereophotogrammetric system. Accelerometer data were collected using wearable acceleration sensors positioned on the prosthesis shank, located distal to the prosthetic knee, and on the subject's residual limb, located proximal to the prosthetic

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knee. From the accelerometer data, we computed the root mean square value of jerk about toe-off and in terminal swing. Jerk was calculated by differentiating the raw acceleration signal, and was used as a measure of gait smoothness. Average parameter values for each prosthesis were calculated and statistically compared using repeated measures ANOVAs and pairwise comparisons performed using the least significant difference test as per all other variables analyzed in the study.

RESULTS

Indoor Track Tests: Comparing Metabolic Cost Across Knee Prostheses

Metabolic cost during steady-state walking at a self-selected, comfortable speed was significantly different across the three tested knees ($P = 0.029$). The rate of oxygen consumption for each of the eight subjects and each of the three tested prostheses are shown in Figure 2. For six out of eight subjects, the rate of oxygen consumption when users wore the Rheo knee was lower compared with the Mauch. Statistical analysis showed a significant decrease in rate of oxygen consumption for the Rheo compared with the Mauch ($P = 0.009$) with an average decrease equal to 5% across the eight subjects. For six out of eight subjects, lower oxygen consumption rates were found for the Rheo as compared with the C-leg with an average decrease equal to 3%. A pairwise comparison revealed a trend, but no statistical significance, between the rate of oxygen consumption for the Rheo and the C-leg ($P = 0.092$). An average difference of 2% was shown between the rate of oxygen consumption for the C-leg and the Mauch, but the difference was not statistically significant ($P = 0.250$).

Laboratory Tests: Comparing Gait Data Across Knee Prostheses

Analysis of temporal parameters recorded in the laboratory setting showed few differences in the timing of ambulation among the three prosthetic knees. The results are summarized in Table 2. The walking speeds across the three prostheses were not statistically different. No statistical difference was demonstrated for the temporal parameters of the unaffected side. For the affected side, only step time demonstrated a statically significant difference among the three knees. Pairwise comparisons showed that the Rheo was associated with generally longer step times than the C-leg and the Mauch.

Analysis of kinematics and kinetics showed that significant differences mark the three knee technologies investigated in the study. An overview of kinematic and kinetic data for hip, knee, and ankle joints are shown in Figure 3. Average data across all the trials and subjects are shown for the affected side for each of the three knees. Differences are suggested by visual inspection of the plots. Conversely, the kinematics and kinetics of the unaffected side (not shown) did not demonstrate any obvious differences among the three knees. Consequently, statistical analysis was focused on parameters derived from hip, knee, and ankle data of the affected side (i.e., angle, torque, and power trajectories).

Results for the comparison of hip biomechanics across the investigated technologies are shown in Table 3. Differences in hip mechanics across the three prosthetic knees were characterized by statistically comparing the following parameters: peak hip extension angle about toe-off, peak hip flexion torque during terminal stance, peak hip extension torque during terminal swing, peak hip power generation during early stance, peak hip power absorption during mid to terminal stance, peak hip power generation about toe-off, hip work during stance, and hip work during swing. Compared with the C-leg and Rheo knees, the Mauch was found to be marked by greater stance period negative hip work production, greater swing period positive hip work production, larger peak hip flexion torque at terminal stance, and larger peak hip power generation at toe-off. Finally, the magnetorheological-based

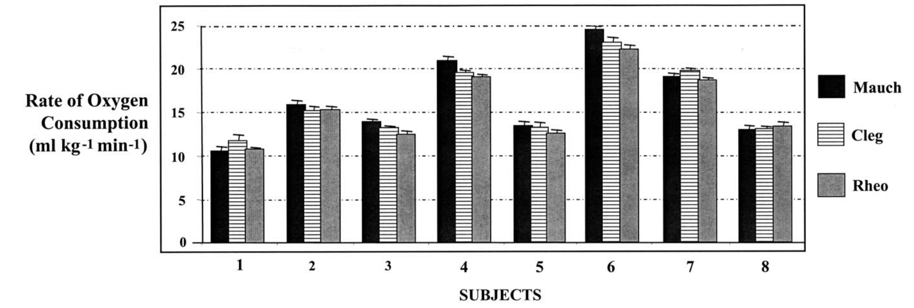

FIGURE 2 *Rate of oxygen consumption for the three prosthetic knees: the mechanically passive hydraulic-based Mauch SNS, the variable-damping hydraulic-based C-leg, and the variable-damping magnetorheological-based Rheo. Error bars represent the standard deviation for each measure.*

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TABLE 2 Temporal parameters for the three prosthetic knees

	Average Values			*P* Values			
Temporal parameter	**Mauch**	**Cleg**	**Rheo**	**ANOVA**	**Mauch *vs.* Cleg**	**Mauch *vs.* Rheo**	**Cleg *vs.* Rheo**
Walking speed (m/s)	1.20	1.18	1.14	NS			
Affected side:							
Step time (s)	0.66	0.65	0.69	**0.019**	NS	**0.038**	**0.007**
Step length (m)	0.76	0.74	0.75	NS			
Single support (s)	0.42	0.42	0.43	NS			
Double support (s)	0.30	0.29	0.32	NS			
Unaffected side:							
Step time (s)	0.57	0.58	0.59	NS			
Step length (m)	0.71	0.71	0.69	NS			
Single support (s)	0.51	0.52	0.53	NS			
Double support (s)	0.31	0.31	0.33	NS			

The average value over eight subjects is shown for each knee. *P* values are shown when smaller than 10%, that is, when at least a trend was identified. Otherwise the differences were considered not significant (NS). *P* values smaller than 5% are in bold type.

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Rheo showed a lower peak hip extension torque during late swing compared with the hydraulic knees, the Mauch and the C-leg (see Table 3).

Results for the comparison of prosthetic knee biomechanics across the investigated technologies are shown in Table 4. Differences in knee mechanics across the three prosthetic knees were characterized by statistically comparing the following parameters: peak knee extension angle at terminal swing, peak knee angular velocity about toe-off, peak knee torque during early stance, peak knee flexion torque at mid to terminal stance, peak knee extension torque about toe-off, peak knee flexion torque at terminal swing, and peak knee power absorption about toe-off. Compared with the Mauch and Rheo knees, peak knee extension angle during terminal swing was found to be significantly larger for the C-leg. Differences among the three knees were also found for peak knee angular velocity about toe-off. The C-leg had a significantly lower angular velocity compared with the Mauch and Rheo knees. Still further, analysis of peak knee torques and powers during the stance period demonstrated significant differences across the three knees, thus suggesting different stance phase control behaviors for the knees investigated. The mechanically passive Mauch had significantly higher values for peak knee extension torque and peak knee power absorption about toe-off compared with the variable-damping C-leg and Rheo knees. Finally, peak knee flexion torque during terminal swing was significantly lower for the magnetorheological-based Rheo compared with the hydraulic-based Mauch and C-leg knees.

Results for the comparison of prosthetic ankle biomechanics across the investigated technologies are shown in Table 5. Differences in ankle mechanics across the three prosthetic knees were characterized by statistically comparing the following parameters: peak ankle plantar flexion angle during early stance, peak ankle dorsiflexion angle during mid to terminal stance, peak ankle plantar flexion torque about 30% of the gait cycle, peak ankle power absorption at midstance, and peak prosthetic foot compression during early stance. Compared with the C-leg, the Mauch and Rheo had significantly greater peak ankle plantar flexion angles and peak foot compressions during early stance. In addition, compared with the C-leg, the Mauch and Rheo had significantly lower peak ankle dorsiflexion angles during mid to terminal stance, as well as lower peak ankle plantar flexion torques about 30% of the gait cycle.

Laboratory Tests: Comparing EMG and Accelerometer Data Across Knee Prostheses

EMG and accelerometer data were studied to explore the potential of wearable technology to capture biomechanical differences among the three knee technologies and to complement the results derived from the traditional kinematic and kinetic gait measures. Tables 6 and 7 summarize the results of the statistical analyses of EMG and accelerometer data. Comparisons of the RMS value of the EMG recordings from the gluteus medius muscle on the affected side showed significant differences among the three knees investigated in the study. Specifically, the magnetorheological-based Rheo was associated with a lower level of muscular activity compared with the hydraulic-based knees, the Mauch and C-leg. The analysis of accelerometer data showed significant differences among the

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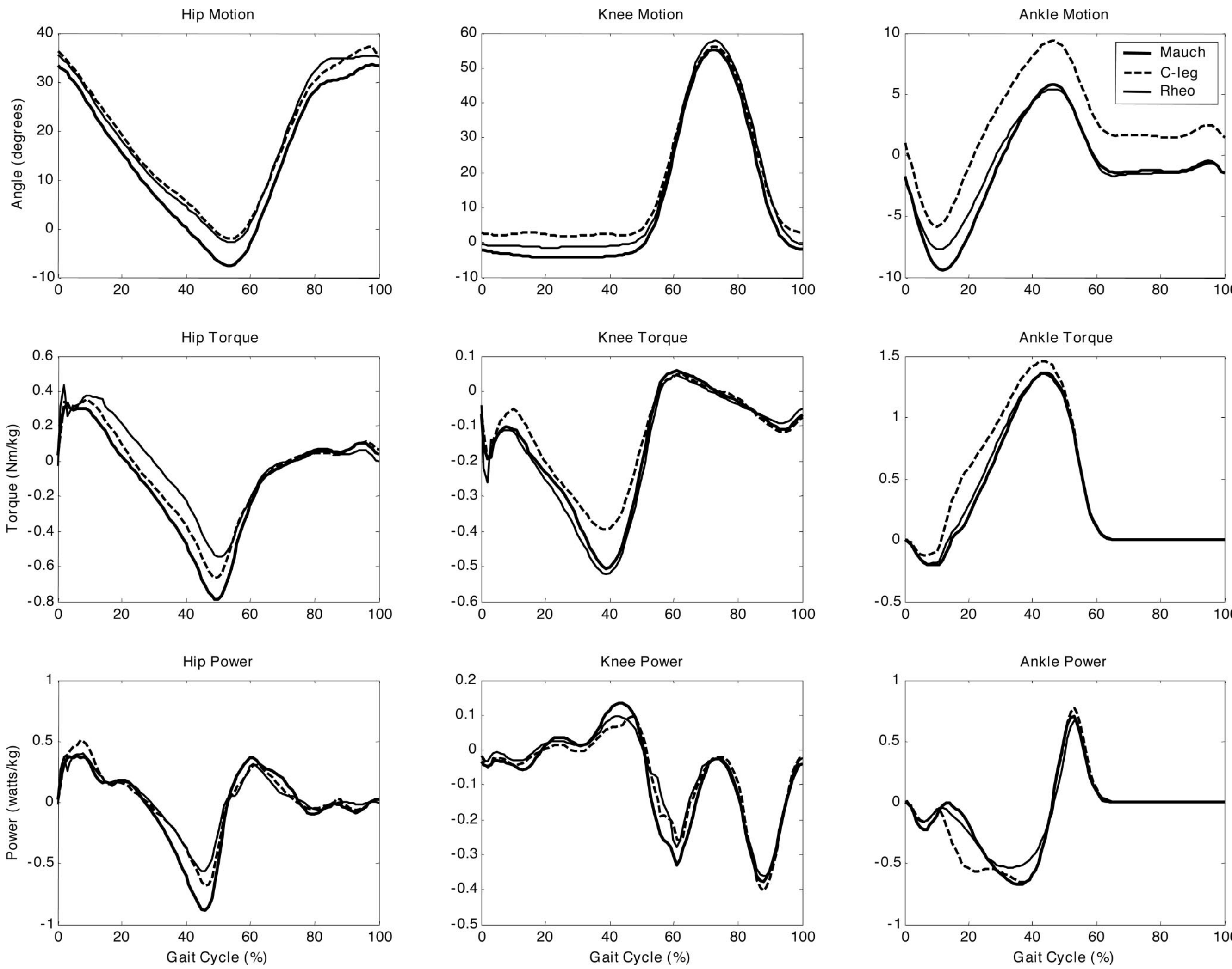

FIGURE 3 *Kinematics and kinetics of gait associated with the three prosthetic knees: the mechanically passive hydraulic-based Mauch SNS, the variable-damping hydraulic-based C-leg, and the variable-damping magnetorheological-based Rheo.*

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three knees about toe-off and in terminal swing. The RMS values of jerk estimated about toe-off from the accelerometer data recorded from the prosthetic shank showed a significant increase for the mechanically passive Mauch compared with the variable-damping Rheo. A trend was also shown with a higher RMS jerk value for the mechanically passive Mauch compared with the variable-damping C-leg. Still further, the RMS values of jerk estimated during terminal swing from the accelerometer data recorded from the thigh demonstrated a significant difference across the three knees investigated in the study. Pairwise comparisons showed a significantly higher RMS jerk value for the mechanically passive Mauch compared with the variable-damping C-leg and Rheo knee prostheses.

DISCUSSION

There is contrasting evidence in the literature concerning whether variable-damping knee prostheses provide a significant advantage over mechanically passive devices specifically at comfortable, self-selected walking speeds. In this investigation, we address this question by comprehensively comparing two distinct variable-damping knee devices, the Otto Bock C-leg and the Össur Rheo, with the mechanically passive Mauch SNS. We hypothesize that variable-damping devices offer an improved metabolic economy of gait compared with mechanically passive designs. Although we find a 2% decrease in metabolic rate for the variable-damping C-leg compared with the mechanically passive Mauch, the difference is not statistically significant (P = 0.250). However, we do find that when using the variable-damping Rheo, energy expenditure decreases by 5% as compared with the Mauch (P = 0.009). We further hypothesize that the distinct torque-producing strategies of the investigated knees, hydraulic *vs.* magnetorheological, result in differences in metabolic gait economy. Our metabolic results support this hypothesis. When

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TABLE 3 Affected-side hip mechanics across the three prosthetic knees

Parameter	Average Values: Mauch	Cleg	Rheo	*P* Values: ANOVA	Mauch *vs.* Cleg	Mauch *vs.* Rheo	Cleg *vs.* Rheo
Peak hip angle in terminal stance (degrees)	−9.1	−3.9	−4.6	0.091	**0.045**	0.077	NS
Peak hip torque in terminal stance (Nm/kg)	−0.945	−0.801	−0.820	**0.043**	**0.021**	**0.041**	NS
Peak hip torque in terminal swing (Nm/kg)	0.163	0.163	0.092	**<0.001**	NS	**<0.001**	**<0.001**
Peak hip power in early stance (watts/kg)	0.67	0.76	0.69	NS			
Peak hip power in mid-to-terminal stance (watts/kg)	−0.97	−0.76	−0.67	0.058	0.090	**0.022**	NS
Peak hip power about toe-off (watts/kg)	0.81	0.63	0.69	**0.015**	**0.005**	**0.040**	NS
Negative hip work during stance (joules/kg)	15.8	10.8	10.3	**0.006**	**0.007**	**0.003**	NS
Negative hip work during swing (joules/kg)	1.8	1.5	1.3	NS			
Positive hip work during stance (joules/kg)	9.9	10.0	9.9	NS			
Positive hip work during swing (joules/kg)	4.7	3.9	3.5	0.071	0.095	**0.027**	NS

Hip angle values are positive for hip flexion and negative for hip extension. Hip torque values are positive for internal hip extension torque and negative for internal hip flexion torque. Hip power values are positive for power generation and negative for power absorption. Torque, power, and work values are normalized by body mass. The average value over eight subjects is shown for each knee. *P* values are shown when smaller than 10%, that is, when a trend was identified. Otherwise the differences were considered not significant (NS). *P* values smaller than 5% are in bold type.

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using the magnetorheological-based Rheo, metabolic rate decreases by 5% compared with the hydraulic-based Mauch ($P = 0.009$) and by 3% compared with the hydraulic-based C-leg (trend; $P = 0.092$).

Additionally, we hypothesize that gait biomechanics are significantly different between variable-damping and mechanically passive prostheses and between hydraulic and magnetorheological-based systems. Our biomechanical results support these hypotheses. We observe several biomechanical advantages for the variable-damping devices compared with the mechanically passive Mauch. These advantages include an enhanced smoothness of gait as indicated by a lower jerk RMS, a decrease in hip work production during stance and swing phases, a lower peak hip flexion moment at terminal stance, and a reduction in peak hip power generation at toe-off. In addition to these biomechanical advantages, the magnetorheological-based Rheo offers an improved prosthetic foot-ground interaction and swing phase hip biomechanics. Compared with the C-leg, the Rheo knee allows for increased prosthetic heel compression, or energy storage, and a reduction in peak hip extension torque during terminal swing.

Finally, we hypothesize that differences in movement and muscle activation patterns associated with the three knees can be captured by means of wearable sensors. Our findings support this hypothesis. For the EMG measures, the magnetorheological-based Rheo is associated with a lower level of muscular activity in the gluteus medius muscle compared with the hydraulic-based knees,

TABLE 4 Affected-side knee mechanics across the three prosthetic knees

Parameter	Average Values: Mauch	Cleg	Rheo	*P* Values: ANOVA	Mauch *vs.* Cleg	Mauch *vs.* Rheo	Cleg *vs.* Rheo
Peak knee angle in terminal swing (degrees)	−2.4	2.7	−0.8	**0.004**	**0.001**	NS	**0.017**
Peak knee angular velocity about toe-off (degrees/s)	384	353	395	**<0.001**	**0.001**	NS	**<0.001**
Peak knee torque in early stance (Nm/kg)	−0.015	0.003	−0.021	NS			
Peak knee torque in mid-to-terminal stance (Nm/kg)	−0.539	−0.427	−0.565	**0.031**	**0.037**	NS	**0.014**
Peak knee torque about toe-off (Nm/kg)	0.127	0.093	0.076	**0.005**	**0.023**	**0.002**	NS
Peak knee torque in terminal swing (Nm/kg)	−0.122	−0.125	−0.097	**0.003**	NS	**0.004**	**0.002**
Peak knee power about toe-off (watts/kg)	−0.72	−0.51	−0.45	**0.001**	**0.004**	**<0.001**	NS

Knee angle values are positive for knee flexion and negative for knee (hyper-)extension. Knee torque values are positive for internal knee extension torque and negative for internal knee flexion torque. Knee power values are positive for power generation and negative for power absorption. Torque, power, and work values are normalized by body mass. The average value over eight subjects is shown for each knee. *P* values are shown when smaller than 10%, that is, when a trend was identified. Otherwise the differences were considered not significant (NS). *P* values smaller than 5% are in bold type.

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TABLE 5 Affected-side ankle mechanics across the three prosthetic knees

Parameter	Average Values: Mauch	Cleg	Rheo	*P* Values: ANOVA	Mauch *vs.* Cleg	Mauch *vs.* Rheo	Cleg *vs.* Rheo
Peak ankle angle in early stance (degrees)	−9.6	−6.2	−8.0	**0.014**	**0.004**	NS	0.077
Peak ankle angle in mid-to-terminal stance (degrees)	6.1	9.8	5.9	**0.019**	**0.015**	NS	**0.012**
Peak ankle torque about 30% of gait cycle (Nm/kg)	0.701	0.978	0.785	**0.002**	**<0.001**	NS	**0.009**
Peak ankle power in mid-stance (watts/kg)	−0.84	−0.79	−0.76	NS			
Foot compression in early stance (mm)	23.8	17.1	24.0	**0.039**	**0.028**	NS	**0.024**

Ankle angle values are positive for ankle dorsiflexion and negative for ankle plantar flexion. Ankle torque values are positive for internal ankle plantar flexion torque and negative for internal ankle dorsiflexion torque. Ankle power values are positive for power generation and negative for power absorption. Torque, power, and work values are normalized by body mass. The average value over eight subjects is shown for each knee. *P* values are shown when smaller than 10%, that is, when a trend was identified. Otherwise the differences were considered not significant (NS). *P* values smaller than 5% are in bold type.

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the C-leg and the Mauch. The analysis of accelerometer data show significant differences among the three knees tested about toe-off and terminal swing. The variable-damping knees generally show lower jerk values about toe-off compared with the mechanically passive Mauch, indicating a smoother transition for the variable-damping devices from stance to swing. Further, the variable-damping knees generally show lower jerk values during terminal swing compared with the Mauch, indicating a smoother transition from swing to stance.

The results of this study indicate that variable-damping knee prostheses have significant advantages over mechanically passive designs for unilateral transfemoral amputees walking at self-selected ambulatory speeds. Moreover, the study results suggest that a magnetorheological-based system may have advantages over hydraulic-based designs. In the following sections, we discuss various biomechanical mechanisms for the observed metabolic differences between the three knee prostheses.

Biomechanical Mechanisms for Metabolic Cost Differences: Rheo *vs.* Mauch

By actively modulating knee joint damping, the Rheo offers increased stance stability compared with the mechanically passive Mauch.[8] Consequently, since the Mauch is mechanically passive and cannot actively modulate joint damping, the knee is typically aligned anteriorly (i.e., the ground reaction force is anterior to the knee center of rotation when the subject is in a quiet stance position) so as to limit the tendency of the knee to flex at heel strike when weight is applied to the prosthesis. In distinction, the Rheo knee is typically aligned in a neutral or posterior manner, and early stance stability is then achieved by increasing knee joint damping. Although the anterior alignment makes the Mauch knee more stable during early stance, that alignment makes rapid knee flexion during preswing more difficult compared with the posteriorly aligned, variable-damping Rheo prosthesis.

Numerous biomechanical differences were observed from the analysis of the laboratory data that seem to be related to the different prosthetic

TABLE 6 Root-mean-square of EMG data over one gait cycle for the gluteus maximus and gluteus medius muscles of the affected side

Muscle	Average Values: Mauch	Cleg	Rheo	*P* Values: ANOVA	Mauch *vs.* Cleg	Mauch *vs.* Rheo	Cleg *vs.* Rheo
Gluteus maximus (RMS, μV)	38.8	56.4	46.0	NS			
Gluteus medius (RMS, μV)	54.5	55.7	33.6	**0.018**	NS	**0.015**	**0.011**

The average value over eight subjects is shown for each knee. *P* values are shown when smaller than 10%, that is, when a trend was identified. Otherwise the differences were considered not significant (NS).*P* values smaller than 5% are in bold type.

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TABLE 7 Root-mean-square jerk parameters derived from accelerometer data for the affected thigh and shank

	Average Values			*P* Values			
Parameter	Mauch	Cleg	Rheo	ANOVA	Mauch *vs.* Cleg	Mauch *vs.* Rheo	Cleg *vs.* Rheo
RMS of thigh jerk about toe-off (m/s^3)	2.358	2.154	2.024	NS			
RMS of shank jerk about toe-off (m/s^3)	1.962	1.541	1.388	**0.037**	0.057	**0.014**	NS
RMS of thigh jerk in terminal swing (m/s^3)	3.340	1.769	2.291	**0.009**	**0.004**	**0.040**	NS
RMS of shank jerk in terminal swing (m/s^3)	2.886	2.231	2.865	NS			

The average value over eight subjects is shown for each knee. *P* values are shown when smaller than 10%, that is, when a trend was identified. Otherwise the differences were considered not significant (NS). *P* values smaller than 5% are in bold type.

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alignment and damping control strategies of the Rheo and Mauch knee prostheses. When subjects wore the Mauch, they utilized a hip control strategy marked by exaggerated hip movement components compared with the Rheo as shown by larger peak hip torque during terminal stance, peak hip power during mid-to-terminal stance, peak hip power about toe-off, and negative hip work during stance. In addition, a trend was observed indicating a larger peak hip extension angle during terminal stance for the Mauch compared with the Rheo. Still further, the larger peak knee moment and peak knee power absorption about toe-off found for the Mauch compared with the Rheo seems to reflect the relative ease of rotation during preswing for the Rheo.

In addition to differences during the stance period of walking, biomechanical observations between the two knees suggested that the Rheo required less effort during the swing phase compared with the Mauch. A significantly larger peak hip flexion moment during terminal swing and a greater swing phase positive work production was found for the Mauch compared with the Rheo. This seems to point to an exaggerated hip control needed when subjects wore the Mauch compared with Rheo. Also, a larger peak knee moment during terminal swing was found for the Mauch compared with the Rheo, further suggesting that the Rheo required less effort during the swing phase of walking.

The wearable sensors employed in the study seem to have captured at least some aspects of the aforementioned differences between the Mauch and Rheo knees. A larger RMS value marked the EMG recordings from the gluteus medius muscle when subjects wore the Mauch compared with the Rheo. Additionally, accelerometer data from the affected shank captured the ease of initiating swing for the Rheo knee compared with the Mauch. The RMS value of jerk about toe-off was larger for the Mauch compared with the Rheo knee. Additionally, accelerometer data from the affected thigh captured a greater swing leg smoothness for the Rheo compared with the Mauch. The RMS value of jerk about terminal swing was significantly larger for the Mauch compared with the Rheo.

These biomechanical differences observed in the laboratory setting via analysis of data gathered using a video camera–based system, as well as wearable sensors, seem to account for the differences in metabolic cost between the Mauch and Rheo knee prostheses. It is worth emphasizing that an average difference of 5% in the rate of oxygen uptake was demonstrated, thus suggesting that the combination of variable-damping and magnetorheological technologies leads to a significant advantage over mechanically passive, hydraulic-based knees. The observed difference in rate of oxygen uptake corresponds to about 20% of the difference between above-knee amputees and healthy adults.[21] Thus, the Rheo knee prosthesis fills in a significant percentage of the gap between the rate of oxygen uptake observed for the Mauch prosthetic knee and the one expected for a control group of healthy adults. Furthermore, the improvement achieved with the Rheo prosthetic knee compared with the Mauch knee is similar in magnitude to the one observed for orthotic interventions and aerobic conditioning exercise protocols in hemiplegic individuals.[21] Based on this observation, we think that it is reasonable to hypothesize that the Rheo prosthetic knee has a significant impact on mobility in above-knee amputees compared with the Mauch prosthetic knee. Future studies performed in the field (i.e., the home and the community settings) by relying on wearable technology seem to be the most appropriate way to investigate this hypothesis.[23,24]

Biomechanical Mechanisms for Metabolic Cost Differences: Rheo *vs.* C-leg

Both the Rheo and C-leg are variable-damping prosthetic knees and thus provide stability during early stance. Due to their posterior alignment, the variable-damping Rheo and C-leg do not require the use of an exaggerated hip strategy to facilitate

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knee flexion during preswing as observed with the anteriorly aligned Mauch knee. However, differences do exist between these two knee prostheses. The Rheo and C-leg are distinct in their torque producing strategies, or magnetorheological *vs.* hydraulic, respectively. In addition, the knees are distinct in their peak extension angle. Upon full extension, the C-leg assumes a slightly flexed knee posture whereas the Rheo is set at zero flexion (see Table 4). The different torque producing strategies seem to influence hip behavior during swing. The Rheo knee is marked by a smaller peak hip moment during terminal swing and a lower gluteus medius muscle activity compared with the C-leg, suggesting an easier swing control for the Rheo. Still further, the fact that the C-leg knee is set in slight flexion seems to influence prosthetic foot-ground interactions during early stance. We find that the C-leg has a significantly smaller heel compression, or prosthetic foot energy storage, compared with the Rheo. The results of this study suggest that differences in hip swing phase behavior and prosthetic foot energy storage are contributing factors for the observed difference in walking metabolism between the Rheo and C-leg knee prostheses.

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Concluding Remarks

The results of this study indicate that variable-damping knee prostheses have significant advantages over mechanically passive designs for unilateral transfemoral amputees walking at self-selected speeds. For the investigated variable-damping devices, the Rheo and C-leg knee prostheses, we observe biomechanical advantages over the mechanically passive Mauch. These advantages include an enhanced smoothness of gait, a decrease in hip work, a lower peak hip flexion moment at terminal stance, and a reduction in peak hip power generation at toe-off. The study results further suggest that the magnetorheological-based Rheo may have advantages over the hydraulic-based C-leg. When using the Rheo, metabolic rate decreases by 5% compared with the Mauch and by 3% compared with the C-leg. In distinction, when using the C-leg, metabolic rate decreases by 2% compared with the Mauch but the difference is not statistically significant. We consider these differences to be clinically relevant and anticipate a significant impact of such differences on mobility. It is our hope that this work will lead to further studies linking prosthetic design to clinical outcomes, resulting in an even wider range of locomotory performance advantages for contemporary prostheses.

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REFERENCES

1. Flowers WC: *A man-interactive simulator system for above-knee prostheses studies* [PhD Thesis]: Massachusetts Institute of Technology; 1972
2. Grimes DL, Flowers WC, Donath M: Feasibility of an active control scheme for above knee prosthetics. *J Biomech Eng–Transactions of the ASME* 1977;99:215–21
3. Darling D: Automatic damping profile optimization for computer controlled above-knee prostheses [Masters Thesis]: Massachusetts Institute of Technology; 1978
4. Popovic D, Schwirtlich L (eds.): Belgrade active A/K prosthesis. Amsterdam, The Netherlands: Excerpta Medica; 1988. de Vries (ed.). J *Electrophysiological Kinesiology, Intern.* Congress Ser. No. 804
5. James K, Stein RB, Rolf R, Tepavic D: Active suspension above-knee prosthesis. Paper presented at: Proceedings of the 6th International Conference on Biomedical Engineering; December, 6–8, 1990; Singapore
6. Kitayama I, Nakagawa N, Amemori K: A microcomputer controlled intelligent A/K prosthesis. Paper presented at: Proceedings of the 7th World Congress of the International Society for Prosthetics and Orthotics, 1992; Chicago
7. Zahedi S: The results of the field trial of the Endolite Intelligent Prosthesis. Paper presented at: XII International Congress of INTERBOR; September 22–25, 1993
8. Herr H, Wilkenfeld A: User-adaptive control of a magnetorheological prosthetic knee. *Ind Rob* 2003;30:42–55
9. Marks LJ, Michael JW: Science, medicine, and the future. Artificial limbs. *Br Med J* 2001;323:732–35
10. Michael JW: Modern prosthetic knee mechanisms. *Clin Orthop Relat Res* 1999;361:39–47
11. Popovic D, Sinkjaer T: *Control of Movement for the Physically Disabled.* London: Springer-Verlag; 2000
12. Datta D, Howitt J: Conventional versus microchip controlled pneumatic swing phase control for transfemoral amputees: user's verdict. *Prosthet Orthot Int* 1998;22:129–35
13. Kirker S, Keymer S, Talbot J, Lachmann S: An assessment of the intelligent prosthesis. *Clin Rehabil* 1996;10:267–73
14. Taylor MB, Clark E, Offord EA, Baxter C: A comparison of energy expenditure by a high level trans-femoral amputee using the Intelligent Prosthesis and conventionally damped prosthetic limbs. *Prosthet Orthot Int* 1996;20:116–21
15. Buckley JG, Spence WD, Solomonidis SE: Energy cost of walking: comparison of "Intelligent Prosthesis" with conventional mechamisms. *Arch Phys Med Rehabil* 1997;78: 330–33
16. Kastner J, Nimmervoll R, Kristen H, Wagner P: Was kann das C-leg? Ganganalytisher Vergleich von C-leg, 3R45and 3R80. *Med Orth Tech* 1999;119:131–37
17. Schmalz T, Blumentritt S, Jarasch R: Energy expenditure and biomechanical characteristics of lower limb amputee gait: the influence of prosthetic alignment and different prosthetic components. *Gait Posture* 2002;16:255–63
18. Flynn K, Alligood E: Computerized lower limb prostheses. VA Technology Assessment Program Short Report; 2000, pp. 1–11

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19. State of Washington DoLaI. Microprocessor-controlled prosthetic knees; 2002
20. Martin CW: A review of microprocessor-controlled knee prostheses: Worker's compensation board of British Columbia; 2003
21. Waters RL, Mulroy S: The energy expenditure of normal and pathologic gait. *Gait Posture* 1999;9:207–31
22. Jolly MR, Bender JW, Carlson DJ: Properties and applications of commercial magnetorheological fluids. Paper presented at: SPIE 5th Annual Int Symposium on Smart Structures and Materials; March 15, 1998; San Diego, CA
23. Bonato P: Advances in wearable technology and applications in physical medicine and rehabilitation. *J Neuroengineering Rehabil* 2005;2:2
24. Bussmann JBJ, Culhane KM, Horemans HLD, Lyons GM, Stam HJ: Validity of the prosthetic activity monitor to assess the duration and spatio-temporal characteristics of prosthetic walking. *IEEE Trans Neural Syst Rehabil Eng* 2004;12:379–86
25. Dempster W: Space Requirements of the Seated Operator. WADC Technical Report: Wright-Patterson Air Force Base, Ohio; 1955, pp. 55–159
26. Perry J: *Gait Analysis: Normal and Pathological Function.* Slack Inc, 1992
27. Dawson B, Trapp RG: *Basic & Clinical Biostatistics*. McGraw-Hill, 2001

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