Feature

User-adaptive control of a magnetorheological prosthetic knee

Hugh Herr and
Ari Wilkenfeld

The authors

Hugh Herr is based at Artificial Intelligence Lab, and Harvard/MIT Division of Health Sciences and Technology, MIT, Cambridge, MA, USA and Department of Physical Medicine and Rehabilitation, Harvard Medical School, Spaulding Rehabilitation Hospital, Boston, MA, USA.
Ari Wilkenfeld is based at Artificial Intelligence Lab, MIT, Cambridge, MA, USA.

Keywords

Control, Medical

Abstract

A magnetorheological knee prosthesis is presented that automatically adapts knee damping to the gait of the amputee using only local sensing of knee force, torque, and position. To assess the clinical effects of the user-adaptive knee prosthesis, kinematic gait data were collected on four unilateral trans-femoral amputees. Using the user-adaptive knee and a conventional, non-adaptive knee, gait kinematics were evaluated on both affected and unaffected sides. Results were compared to the kinematics of 12 age, weight and height matched normals. We find that the user-adaptive knee successfully controls early stance damping, enabling amputee to undergo biologically-realistic, early stance knee flexion. These results indicate that a user-adaptive control scheme and local mechanical sensing are all that is required for amputees to walk with an increased level of biological realism compared to mechanically passive prosthetic systems.

Electronic access

The research register for this journal is available at
http://www.emeraldinsight.com/researchregister

The current issue and full text archive of this journal is available at
http://www.emeraldinsight.com/0143-991X.htm

Emerald

Industrial Robot: An International Journal
Volume 30 · Number 1 · 2003 · pp. 42–55
© MCB UP Limited · ISSN 0143-991X
DOI 10.1108/01439910310457706

Introduction

We know from early Roman mosaics that trans-femoral prostheses have been used during much of recorded history. The earliest involved a simple stick or peg leg. Later, a hinge was introduced to allow the knee to bend during the swing phase of walking. During the Napoleonic wars, Lord Uxbridge, Wellington's cavalry officer at Waterloo, wore a hinge-type trans-femoral prosthesis that even dorsiflexed the foot, as the knee flexed, to reduce stumbling during the swing phase. Although the hinge-type design was an improvement over simple peg leg constructions, the design failed to offer amputees adequate stability. Modern knees, developed after World War II, improved upon the concept of a simple hinge joint by adding hydraulic cylinders capable of damping knee rotations. Although important mechanically-passive knees were advanced during the post WWII era (1945-1970), it was not until the 1970s that researchers began developing highly adaptive, electronically-controlled prosthetic knees.

Electronic knees use some form of computational intelligence to control the resistive torque or damping about the knee joint and offer several advantages over mechanically passive designs (Popović and Sinkjaer, 2000). Electronic knees can be programmed to detect stumbles and other pathological behaviors and react appropriately. Using sensory information, they can provide for a more natural gait by discriminating between early and late stance, enabling amputees to flex their knee just after heel strike. This feature of normal walking is important for overall leg shock absorption and is not achievable with most mechanically passive prostheses (Aeyels *et al.*, 1992; Gard, 1999; Peeraer *et al.*, 1988). Electronic knees can also offer different levels of damping during the swing phase and optimize damping levels at different walking speeds. Electronic knees can even detect stairs, sitting down,

We thank the members of Spaulding Gait Laboratory, Spaulding Hospital, Boston, MA, for their kind assistance throughout the study. Additionally, we thank the members of Next Step O&P, Manchester, NH, for their helpful assistance in the area of prosthetics. We also thank the experimental participants for their time and patience. This research was supported by Össur, a manufacturer of leg prostheses.

and other non-standard gait behaviors and respond appropriately.

Several research groups have been involved in the design of prototype knee controllers for use in the laboratory. Beginning in the 1970s, Woodie Flowers and his students at the Massachusetts Institute of Technology worked on a variety of microprocessor controlled knees. Two of his students, Darling (1978) and Grimes (1979) worked on controller designs based on the concept of "echoing" the actions of the sound leg to control the prosthetic leg. Following Flowers' seminal work, Kautz and Seireg (1980) and later Bar *et al.* (1983) also designed a knee based on input from the sound side leg. In addition to these "echo" controllers, laboratory researchers also experimented with electromyographic signals in the control of a trans-femoral knee prosthesis. Myers and Moskowitz (1981, 1983) and Triolo and Moskowitz (1982) worked with electromyographic voluntary control of a knee prosthesis, as did Peeraer *et al.* (1989) and later Aeyels *et al.* (1995). Academic research not only focused on different sensory modalities but also on novel control strategies. Popović and Kalanović (1993) and Popović *et al.* (1991) worked on using output space Lyapunov tracking for control of an active knee prosthesis while Ju *et al.* (1995) attempted to use "fuzzy logic" for the same purpose.

Motivated by academic research activities, a small number of companies introduced variable-damper electronic knees for clinical use. Prominent among these is the Otto Bock C-leg. The hydraulic C-leg detects knee position, ankle force and torque, and provides adjustable damping for flexion and extension in swing, and additionally offers damping control throughout stance (Dietl and Bargehr, 1997; James *et al.*, 1990; Kastner *et al.* (1998)). Although the C-leg offers several clinical benefits compared to mechanically passive knees, including improvements in gait stability, the system is not user-adaptive. Before the knee can be used, a trained prosthetist must program knee damping levels to the amputee until the prosthesis is comfortable, moves naturally, and is safe. However, these prosthetic adjustments typically are not guided by biological gait data, and therefore, knee damping may not be set to ideal values, resulting in undesirable gait movements. Still further, since knee damping levels are set to fixed values by a prosthetist, the knee cannot adapt properly to a disturbance once the amputee has left the prosthetics facility. When an amputee lifts a suitcase or carries a backpack, for example, knee damping levels should not remain constant but instead should increase to compensate for the added load on the prosthesis.

In this paper, we ask whether a computer-controlled, variable-damper electronic knee, employing only sensory information measured local to the knee axis, can automatically adapt knee damping values to match the amputee's gait requirements, accounting for variations in forward walking speed, user gait styles and body size.
We hypothesize that a user-adaptive control scheme and local mechanical sensing are all that is necessary for amputees to walk with an increased level of biological realism compared to mechanically passive prosthetic systems.
To test this hypothesis, kinematic gait data are collected on four trans-femoral amputees walking at slow, self-selected and fast speeds. For each participant, the user-adaptive electronic knee and a passive, non-adaptive knee prosthesis are tested, and the results are compared to the gait kinematics of twelve age, weight and height matched normals.

Materials and methods

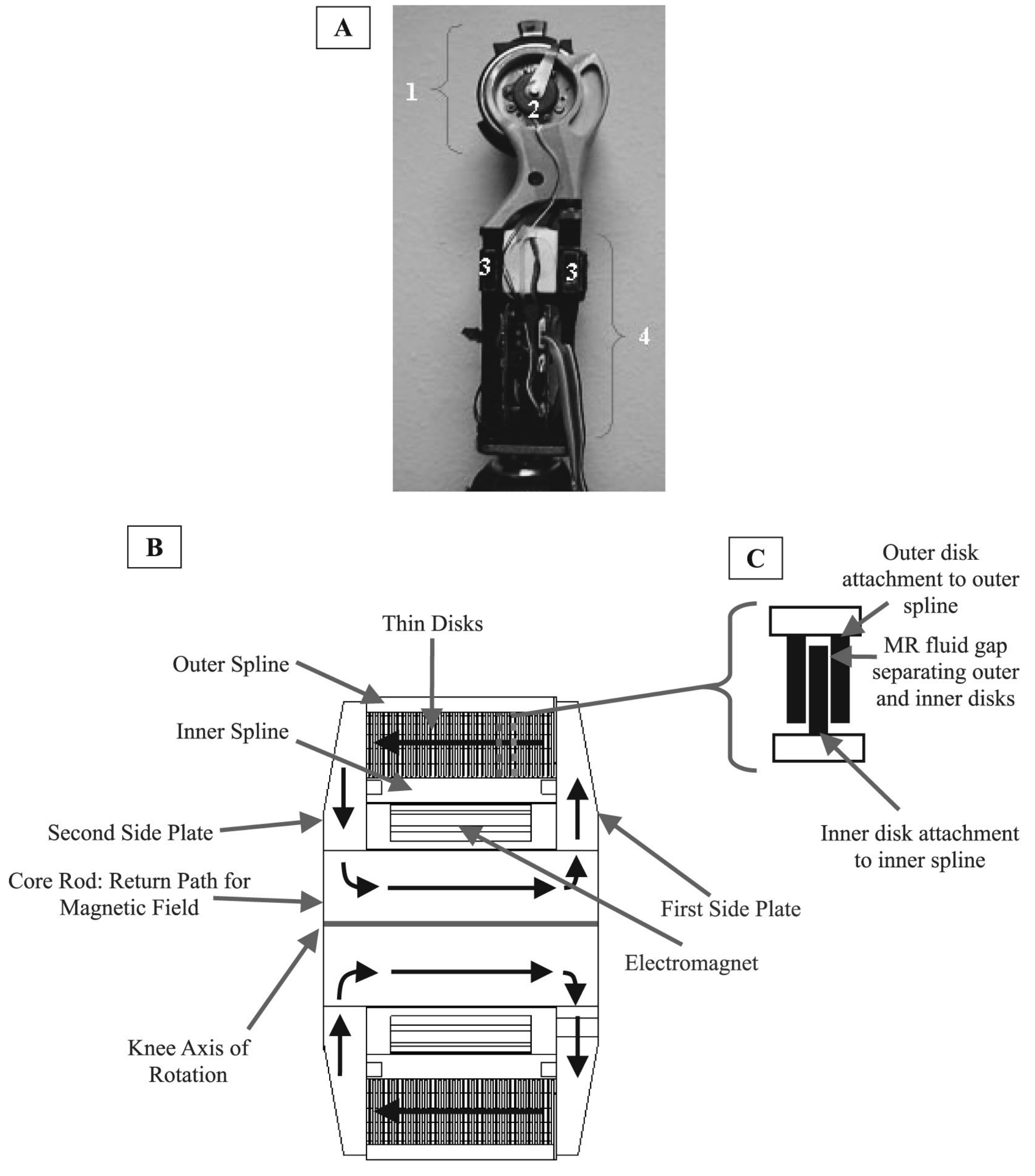
Magnetorheological knee prosthesis: actuator, sensors, microprocessor and battery

To investigate user-adaptive control schemes and their effect on trans-femoral amputee gait, a variable-damper knee prosthesis was developed. The device, shown in Figure 1, is self-contained with (1) actuator, (2) angle sensor, (3) strain gage sensors, and (4) electronic board/battery. The total mass of the prosthesis, including actuator and electronics, is 1.4 kg. In the sections to follow, actuator, sensors, microprocessor and battery are described.

Actuator design

Many brake technologies have been developed for prosthetic knee applications including hydraulic, pneumatic, friction, and magnetorheological (MR) damping strategies (Popović and Sinkjaer, 2000). In this study, MR fluid was used in the shear mode as

Figure 1 A variable-damper knee prosthesis



Note: The prosthesis comprises MR brake (1), potentiometer angle sensor (2), force strain gage sensors (3), and battery and electronic board (4). Current passing through an electromagnet generates a magnetic field that passes through a magnetic core rod, radially outwards through a first side plate, laterally through an interspersed set of thin metal disks, and then radially inwards through a second side plate. The thin disks comprise inner and outer disks where each inner disk is positioned between two outer disk pairs. When the knee rotates, each inner disk moves relative to each outer disk pair. As is shown in Figure 1C, between each inner and outer disk is a thin film of MR fluid (fluid gap ~ 20 microns). As the magnetic field strength increases, micron-sized iron particles within the MR fluid (30% iron loading) form torque-producing chains connecting adjacent disk surfaces. Thus, by controlling electromagnet current, knee damping is controlled. Two dynamic seals hold the MR fluid within the region of interspersed metal disks, and two bearings enable the inner disks to rotate with respect to the outer disks. By increasing electromagnetic current, actuator torque can be increased by three-orders of magnitude from a minimum, zero-current torque level equal to 0.5 newton-meter. The knee's zero-current torque response is due to viscous fluid damping resulting from the shearing of MR fluid between adjacent disk pairs

the primary torque-producing strategy. MR fluid has small iron particles (~ 1 micron) suspended in oil that form torque-producing chains in response to an applied magnetic field. To generate a magnetic field within the MR fluid, the knee brake of this investigation comprised an electromagnet and a magnetic circuit. By varying current in the electromagnet, the magnetic field was controlled within the magnetic circuit and thus the level of MR knee damping.

In Figure 1(B), a coronal section of the knee's magnetic circuit is sketched.

When current was applied to the knee's electromagnet, a magnetic field was generated through a return path centered about the knee's rotary axis. The field then moved radially outwards through a first side plate, laterally through an interspersed set of inner and outer metal disks, and then radially inwards through a second side plate. Inside the knee, each outer and inner disk was shaped like a concentric ring about the knee's axis of rotation. Furthermore, as is shown in Figure 1(C), each outer disk was coupled to an outer spline, and each inner disk was coupled to an inner spline. When the knee was flexed, the inner spline rotated with respect to the outer spline, and therefore each inner disk rotated with respect to two outer disk pairs. Injected between each inner and outer disk was a thin film of MR fluid ($\sim$20 micron gap). When a magnetic field passed through the stack of disks perpendicular to each disk surface, MR damping developed in response to the applied field. MR chains developed within the fluid, connecting each lower disk surface to an adjacent upper disk surface. These chains further enhanced the required torque necessary to rotate the knee, or to shear a lower disk surface relative to an upper disk surface. For a more detailed description of the knee actuator technology, see Deffenbaugh *et al.* (2001).

Sensors, microprocessor and battery

To control knee resistive torque, the prosthesis of this investigation used only local mechanical sensing of knee position, force and torque. Here the phrase "local sensing" means that all sensors were positioned relatively close to the knee axis (<10 cm), allowing amputee's to employ vertical shock pylon technologies critical to overall prosthesis shock absorption. Angle sensor (2) in Figure 1 (custom built potentiometer, 15 kΩ) measured knee flexion angle. The angle signal was then differentiated in analog circuitry to estimate knee angular velocity. Knee velocity was critical for determining whether the knee was flexing or extending. Axial force sensors (3) in Figure 1 (two aft and two fore strain gages) measured the component of force applied to the knee prosthesis from the ground in the direction of the knee's longitudinal axis (add fore and aft strain gage signals). The axial force measurement was critical for determining whether the prosthetic foot was on or off the ground. The strain gage sensors were also used to measure knee torque (subtract fore from aft strain gage signals). Throughout early stance in walking, when only the heel was loaded, the torque sensor measured a positive flexion moment, denoting that the amputee's load line was posterior to the knee's rotational axis and the knee prosthesis was at risk of buckling. In distinction, during late stance, when only the toe was loaded, the torque sensor measured a negative extension moment, denoting that the load line was anterior to the knee's rotational axis and the prosthesis was not at risk of buckling. As will be described in the next section, the controller changed from high to low damping depending on heel versus toe loading conditions, respectively. For this investigation, a 6812 Motorola microprocessor was used for computation, and four rechargeable lithium ion batteries were employed for power.

MR knee prosthesis: control algorithm

Description of normal, level-ground walking:
To describe how the electronic knee prosthesis was controlled, the basic walking progression must first be explained. There are five distinct phases to a walking gait cycle (Inman, 1981).

(1) Beginning with heel strike, the stance knee begins to flex slightly. This flexion allows for shock absorption upon impact as well as keeping the body's center of gravity at a more constant vertical level throughout stance.
(2) After maximum flexion is reached in the stance knee, the joint begins to extend again, until maximum extension is reached.
(3) During late stance, the knee of the supporting leg begins to flex again in preparation for leaving the ground for swing. This is referred to in the literature as "knee break" or "pre-swing". At this time, the adjacent foot strikes the ground and the body is in "double support mode" (that is to say, both legs are supporting body weight).
(4) As the hip is flexed, and the knee has reached a certain angle in knee break, the leg leaves the ground and the knee continues to flex.
(5) After reaching a maximum flexion angle during swing, the knee begins to extend. After the knee has reached full extension,

Industrial Robot: An International Journal
Volume 30 · Number 1 · 2003 · 42–55

the foot once again is placed on the ground, and the next walking cycle begins.

States and transitional conditions

These basic phases of biological gait suggested the framework of the prosthetic knee controller as a state machine. Each phase (one through five) corresponded to a state. Figure 2 shows a graphical representation of a person moving through a normal gait cycle and the location of each state within that cycle. As was discussed earlier, to determine system state, the onboard sensors measured knee angle, force and torque. Based upon these sensory data, the controller cycled through the state machine as the user moved through each gait cycle. In Figure 3, the conditions that must be satisfied to move from state-to-state are specified for a typical walking cycle.

Within each state, electric current through the knee's electromagnet was controlled such that knee resistive torque was proportional to the square of knee rotational velocity, or

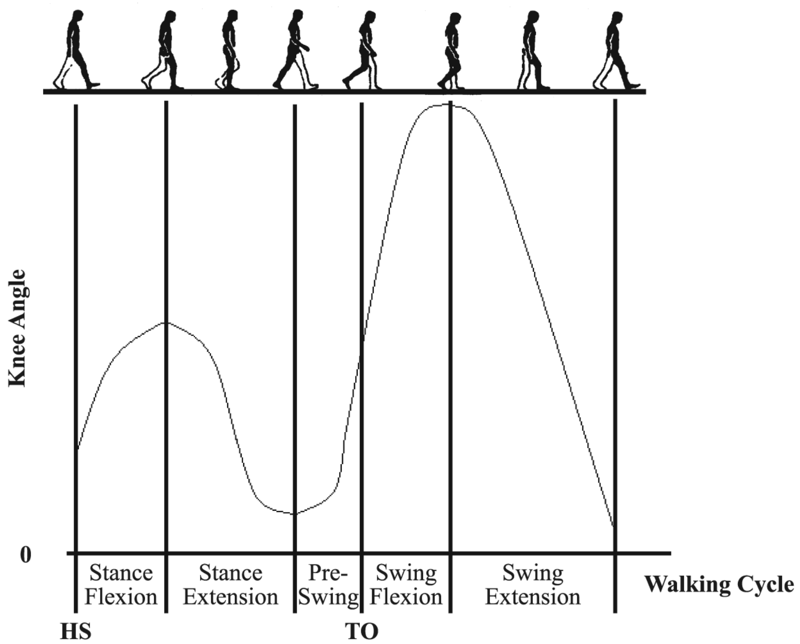
$$\text{Torque} = B(V)^2 \qquad (1)$$

where V is the knee angular velocity from the differentiated angle signal and B is the active knee damping constant. Within a given walking cycle, five distinct values of damping were used corresponding to the five phases or states of the knee controller. Knee damping, B, was only modulated from state-to-state and from cycle-to-cycle but never throughout the duration of a particular gait phase or control state. The objective of the user-adaptive control scheme was to select a knee damping value for each state, B, that would result in an improved trans-femoral gait in terms of biological realism and symmetry between affected and unaffected sides. Specifically, the aim of the controller was to achieve
(1) a biologically realistic maximum flexion angle during the swing phase, and
(2) a biologically realistic early stance flexion-extension cycle critical to overall leg shock absorption. In the next section, control actions for each state are defined. States 1, 2 and 3 are referred to as *stance phase control* and states 4 and 5 as *swing phase control*.

Control algorithm

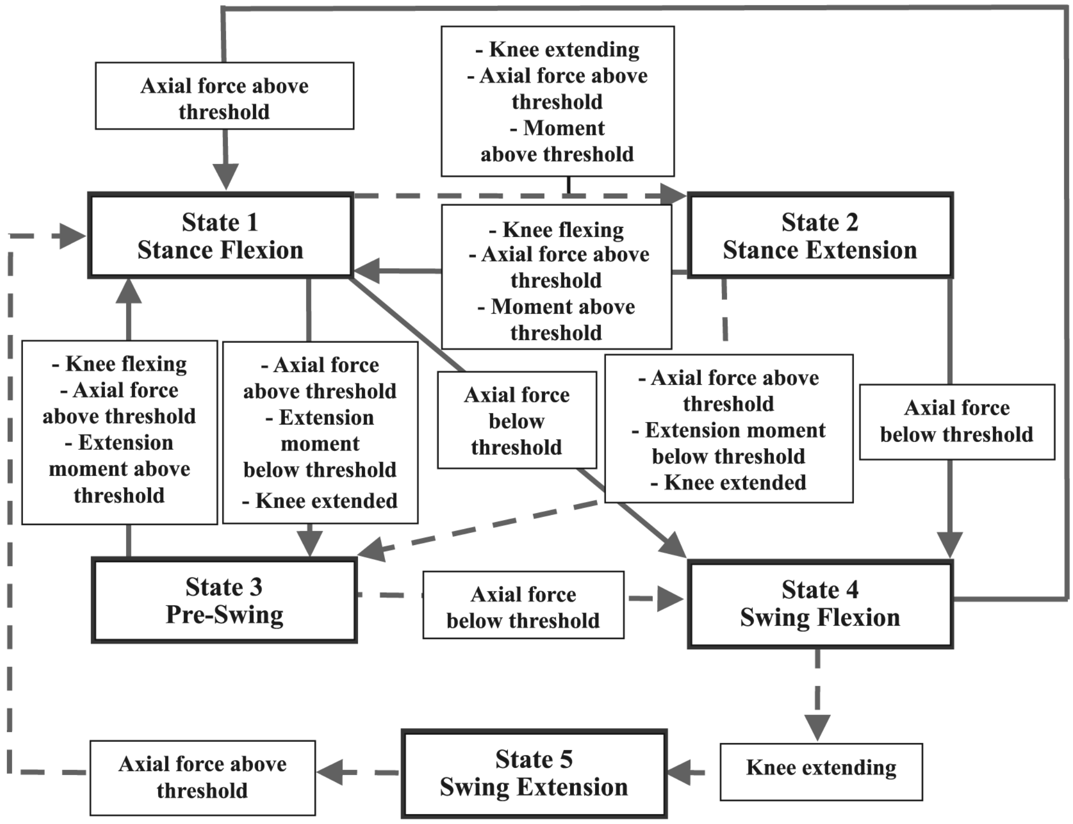
Stance phase control actions: States 1, 2 and 3. In biological gait, the knee first flexes and then extends throughout early to midstance (see Figure 2). In State 1, or Stance Flexion, the prosthetic knee applied a relatively high level of damping (B in equation 1) to inhibit the knee from buckling under the user's weight.

Figure 2 A normal gait cycle is shown schematically with state transitions represented



Note: State 1 and State 2 represent early stance flexion and extension just after heel strike (HS), respectively. State 3, or pre-swing, typically occurs at the end of stance, beginning just after the knee has fully extended and terminating when the foot has left the ground at toe-off (TO). State 4 and State 5 represent periods of knee flexion and extension during the swing phase of walking, respectively

Figure 3 The state machine with state-to-state transitional conditions specified



Note: The state transitions for a typical gait cycle are shown with dashed lines whereas non-typical transitions are specified with solid lines. Starting in State 5, or Swing Extension, State 1 began when the axial force exceeded an onground force threshold equal to 5% of the average stance-period, maximum axial force determined from previous gait cycles. Once in State 1, the controller transitioned to State 2, or Stance Extension, when the knee first began to extend, or when the angular velocity first became negative. By convention, full knee extension corresponded to zero degrees, not 180 degrees. Thus, knee flexion velocity was positive, and knee extension velocity was negative. To transition from State 2 to State 3, two conditions had to be satisfied: 1) the knee had to be fully extended (<5 degrees) for a fixed amount of time (>20ms), and 2) the moment had to fall below a critical moment threshold equal to 30% of the average minimum stance-period moment determined from previous gait cycles. Once in State 3, or Pre-Swing, the controller transitioned to State 4, or Swing Flexion, when the axial force dropped below the onground force threshold. Generally, if at any time during stance (states 1, 2 or 3) the axial force dropped below the onground force threshold, the state machine would transition to State 4. Finally, once in State 4, the controller transitioned to State 5, or Swing Extension, when the knee first began to extend, or when the angular velocity first became negative

The prosthetic knee also applied high damping during the extension period of stance, or State 2, to slow or damp knee extension so that the rotating portion of the knee did not slam against the prosthetic kneecap at full knee extension. In this investigation, the four amputee participants were able to extend their knee by actively extending their hip with the prosthetic foot firmly positioned on the ground.

The degree to which the electronic knee damped flexion and extension was largely dependent on the maximum axial force applied to the prosthesis during stance; the larger the axial loads, the larger the user and the greater was the preferred flexion/extension damping values. In clinical studies, flexion and extension damping values were optimized for amputees of different body size and maximum axial loadings, and these data were then used to establish relationships between early stance damping and axial force sensory information measured during stance.

When an amputee first walked on the knee, State 1 damping was made large so that the knee was safe and did not buckle to exceedingly large flexion angles. In distinction, State 2 damping was made small so that the amputee could successfully extend the knee from a flexed state. However, after the amputee participant took several walking steps ($N = 20$ steps), the damping values (B in equation 1) for States 1 and 2 were made proportional to the average stance-period, peak axial force, F_{max}, or

Industrial Robot: An International Journal
Volume 30 · Number 1 · 2003 · 42–55

$$B = A(F_{\max}) \tag{2}$$

where A is a proportionality constant determined to give biologically realistic early stance flexion-extension dynamics during previously conducted clinical investigations. With this control approach, no user-specific information was programmed into the prosthetic knee. Using sensory information measured local to the knee prosthesis, stance resistances were automatically adapted to the needs of the amputee.

State 3: In State 3, or pre-swing, electromagnet current, or knee active damping B, was set equal to zero. Here the knee's zero-current torque response was due to viscous fluid damping resulting from the shearing of MR fluid between adjacent disk pairs (see Figure 1).

Swing phase control actions: States 4 and 5. During the stance period of a gait cycle, the knee sensors measured a parameter that changed monotonically with locomotory speed. In this investigation, the amount of time the prosthetic foot remained in contact with the ground or foot contact time, was used. As amputee participants walked at increasingly faster speeds, their foot contact time steadily decreased. This basic trend also holds for non-amputees (Wilkenfeld, 2000).

Through an iterative process, the user-adaptive controller determined how swing phase damping should change with foot contact time or walking speed. Stored in the memory of the knee's processor was the full biological range of foot contact time. A person of short stature has, on average, smaller foot contact times compared with a person of tall stature. The full biological range stored in memory included both these extremes.
For this investigation, 0-2 s was more than sufficient to cover the full biological range of foot contact times. This range was partitioned into time slots. A reasonable partition size was 40 ms, giving a total of 50 time slots over the two-second interval. Any one amputee sampled not all but a fraction of the 50 time slots when moving from a slow to a fast walking pace. Since the entire biological range was partitioned, each amputee, independent of height, sampled multiple time slots when accelerating from a slow to a fast speed.

Within each time slot, swing phase flexion damping (B in equation 1) was modulated to control the maximum flexion angle in State 4. When the amputee participants first used the knee prosthesis controlled by this scheme, State 4 damping was set equal to zero (zero electromagnet current) within each time slot. Hence, when the amputee participants took their first step, State 4 damping was minimized, and the knee swung freely throughout the early swing phase. However, for subsequent steps, the controller increased the level of active damping whenever the knee flexed to an angle greater than a fixed target angle. Maximum flexion angle during early swing typically does not exceed 70 degrees in normal walking (Inman, 1981). Hence, to achieve a gait cycle that appears natural or biological, the target angle was set equal to 70 degrees. The amount that damping was increased was proportional to the error between the actual peak flexion angle, measured by the angle sensor (see Figure 1), and the target angle. Increased damping lowered the peak flexion angle in future gait cycles, but only in those time slots or walking speeds for which the amputee had sampled. Knee damping was decreased when the peak flexion angle fell below the target angle for N consecutive walking steps ($N = 20$ gait cycles), ensuring that damping levels would not be unnecessarily high.

As an amputee participant continued to use the prosthesis, sampling a diverse range of walking speeds, knee damping gradually converged within each time slot until peak knee flexion *always* fell below, or close to, the target angle for *all* walking speeds. Hence, once this adaptive scheme was complete, the amputee could rapidly accelerate from a slow to a fast walk all the while sampling different time slots, and therefore, different active damping levels within State 4.

When the amputee participants took their first walking steps, the user-adaptive knee extended from the maximum flexion angle in State 4 to full knee extension in State 5 (extended knee = zero degrees) *with knee damping, B, set equal to zero*. However, for subsequent gait cycles, State 5 damping was gradually increased within the time slots or walking speeds sampled by the amputee participant. The amount of damping increase in State 5 was proportional to the current level of damping in State 4. The level of extension damping during swing was largely dependent on the amount of damping required to constrain the maximum flexion angle to 70 degrees; the larger the flexion damping, the

Industrial Robot: An International Journal
Volume 30 · Number 1 · 2003 · 42–55

larger the user (and lower limb moment of inertia) and the greater was the required State 5 extension damping. In clinical studies, extension damping values were optimized for amputees of different body size and lower limb moments of inertia, and these data were then used to establish a functional relationship between extension and flexion damping.

As the amputee participants continued to use the prosthesis, sampling a diverse range of walking speeds, State 5 damping gradually converged within each time slot (since State 4 damping was convergent). Once the State 4 adaptation scheme was complete, the amputee could rapidly accelerate from a slow to a fast walk all the while sampling different time slots, and therefore, different State 5 damping levels.

Clinical evaluation

Subjects

The clinical evaluation of the electronic knee was conducted in the Gait Laboratory at Spaulding Rehabilitation Hospital, Boston, MA. Protocol approval was provided by the Spaulding Rehabilitation Hospital and Boston University School of Medicine institutional review boards. Moreover, a written informed consent was obtained from each participant before data collection began.

A total of four unilateral trans-femoral amputees (two male, two female) participated in the study (see Table I). Participants were generally in good health and were experienced at prosthesis ambulation; each participant had been an amputee for at least 2 years. Participants were 25-48 years old (mean 39 years), 162-188 cm in height (mean 175 cm), and weighed from 53 to 97 kg (mean 74 kg). A group of 12 unimpaired subjects also participated in the study. Unimpaired subjects spanned a similar age, weight and height range as the four trans-femoral amputee participants.

Amputee participants were asked to commit to three experimental sessions, and unimpaired participants committed to only two sessions. For the amputee participants, the electronic knee and the mechanically passive knee (see Table I) were fitted and aligned during the first session, and during subsequent visits, kinematic gait data were collected at Spaulding Gait Laboratory. For the 12 unimpaired participants, kinematic data were collected during both experimental sessions.

Prosthesis alignment was conducted by a trained prosthetist. It was essential that the same prosthetist align each subject to decrease differences in alignment style. Each knee prosthesis was aligned such that the load line passed posterior to the knee axis (0.5 to 1 mm) when the subject stood with an upright posture. Furthermore, to eliminate variability in ankle-foot systems between subjects, the same foot system was used for each participant. For this investigation, the high energy-return Re-Flex VSP® foot from Össur was attached and aligned to each knee system.

Table I Subject sex, total body mass, height and conventional prosthesis are listed

Subject #	Sex	Body mass (kg)	Height (cm)	Conventional prosthesis
1	M	88	185	Endolite ESK
2	F	53	165	Endolite ESK
3	F	58	162	Otto Bock 3R60
4	M	97	188	Tae Len

Data collection

Kinematic data were measured on both the affected and unaffected sides using an eight-camera VICON 512 system (AMTI Newton, MA). The data were processed at 120 Hz with VICON Workstation (Oxford Metrics, UK) using the standard model of the lower limbs included with the software (Davis *et al.*, 1991; Kadaba *et al.*, 1990; Ramakrishnan *et al.*, 1987). These data, including knee angular position, were then analyzed using MATLAB (Matlab Function Reference, Mathworks, Natick, MA). The eight camera video-based motion analysis system was used to measure the three-dimensional positions of reflective markers placed at various locations across the body. The reflective markers were positioned on bony structures of the pelvis and lower extremities – bilateral anterior superior iliac spines, lateral femoral condyles, lateral malleoli, forefeet, and heels. Additional markers were placed over the sacrum and rigidly attached to wands over the mid-femur and mid-shank.

Test procedure and data analysis

For this investigation, each amputee participant was asked to walk at slow, self-selected and fast walking speeds. The order with which knee systems were evaluated was randomized, and each participant was provided with ample time to acclimatize to each knee before experimentation began. After this habituation, approximately ten walking trials of kinematic data were collected for each steady state speed and knee condition. Kinematic data were also collected on 12 unimpaired participants. Approximately ten walking trials were collected for each steady state speed with a total of 17 different speeds evaluated across a speed range of 0.3-1.8 m/s.

For each participant, experimental trial and control condition, both the maximum knee flexion angle during early stance and the maximum swing flexion angle were computed from the kinematic data. For the amputee participants, both the affected and unaffected sides were analyzed to assess the level of gait symmetry.

Results

Early stance knee flexion

When using the user-adaptive knee prosthesis, all four amputee participants demonstrated early stance knee flexion. At the self-selected walking speed, the maximum flexion angle during early stance was 8 ± 4 degrees. However, no early stance flexion was observed with the mechanically passive prostheses.
In Figure 4, a single walking cycle is plotted showing knee angle (A in degrees), force (B in arbitrary units) and moment (C in arbitrary units) for amputee participant 1 (see Table I) walking at a self-selected speed. These data, measured directly from the electronic knee sensors, show early stance knee flexion with a maximum flexion angle of 9 degrees. Electronic knee state transitions are shown for clarity.

Maximum swing-phase knee flexion

The user-adaptive knee constrained the maximum swing flexion angle to an acceptable biological limit of 70 degrees.
In Figure 5, both the maximum swing flexion angle and the electronic knee damping values are plotted against the number of walking steps taken from a non-adapted prosthetic state. Data are from a single steady state walking speed. During the first ~ 7 walking steps, the maximum swing flexion angle (filled diamonds) is greater than the biological 70 degree threshold. Consequently, the user-adaptive knee controller increased flexion damping (filled squares) until the maximum flexion angle fell below the biological threshold of 70 degrees.

Figure 4 Sensory data from the prosthesis for a single stride showing knee angle (in degrees), force (in arbitrary units) and moment (in arbitrary units) (HS (when the foot first hits the ground) and TO (when the foot leaves the ground for swing) are marked, as are the five states the controller cycles through during each gait cycle. Throughout early stance, the prosthetic knee undergoes a flexion-extension cycle with a maximum flexion angle of 9 degrees)

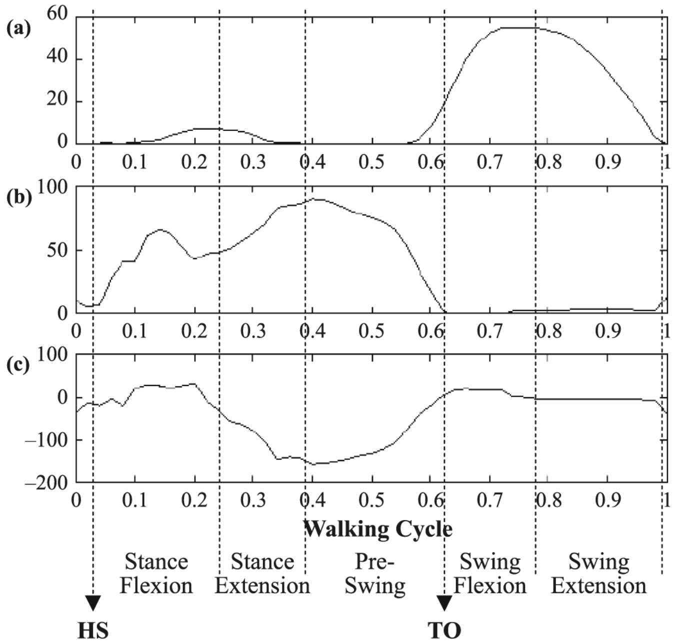
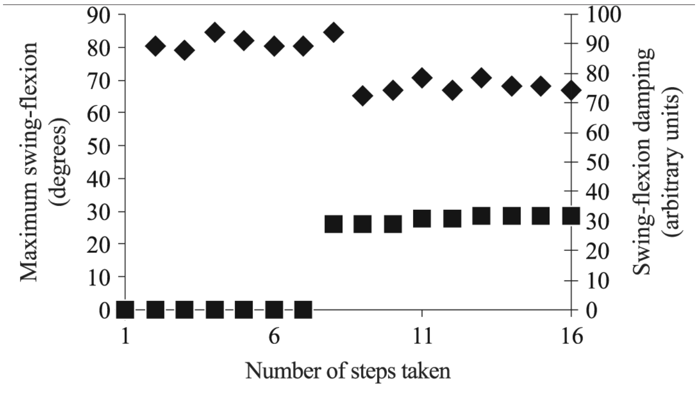


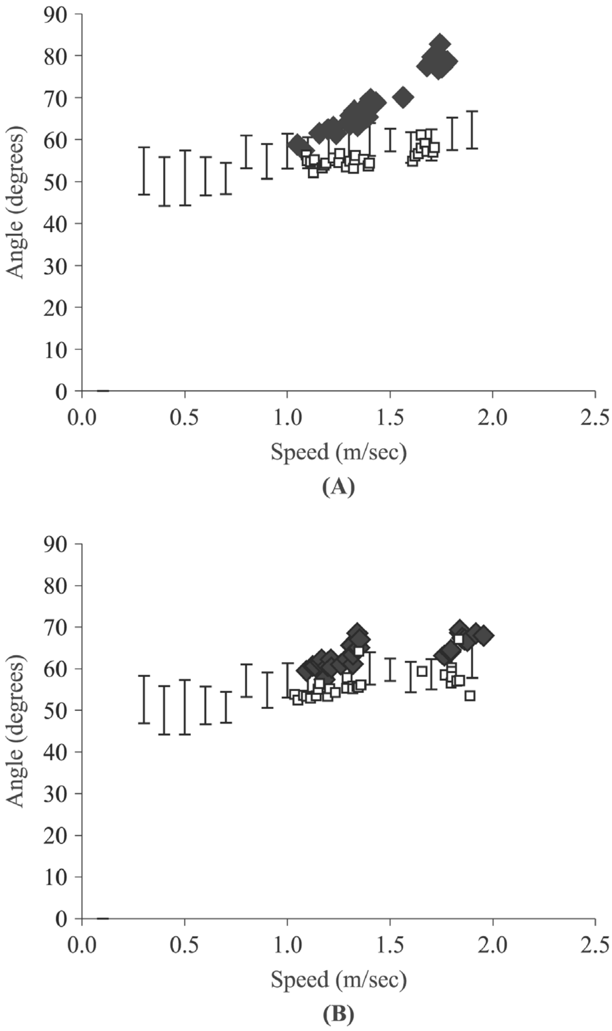
Figure 5 Adaptation of swing-flexion damping at a constant forward walking speed



Note: In State 4, or swing flexion, the objective of the adaptation scheme is to limit the maximum knee angle to less than seventy degrees as this is an acceptable upper limit for biological knees. Swing flexion damping (filled squares) begins at zero for the first several steps. During this time, the maximum knee angle (filled diamonds) is excessively high, triggering the adaptation scheme to increase swing flexion damping until the maximum angle is less than seventy degrees. Although these data were collected at a single forward walking speed, the State 4 adaptation algorithm selects different damping values according to walking speed or time of foot contact range

In Figures 6 and 9, the maximum flexion angle during the swing phase is plotted versus walking speed for subjects 1 through 4, respectively, using the non-adaptive, mechanical knee (plot A, filled diamonds)

Figure 6 Maximum flexion angle during the swing phase versus walking speed for subject 1 (see Table I)



Note: Using the non-adaptive, mechanical knee (plot A, filled diamonds) and the adaptive knee (plot B, filled diamonds). In (A) and (B), the subject's sound side leg is shown (open squares), along with reference data from unimpaired walkers (standard error bars). In (A), the peak angle increases with increasing speed, far exceeding seventy degrees, but in (B) the maximum flexion angle is less than seventy degrees and agrees well with biological data

and the user-adaptive electronic knee (plot B, filled diamonds). In (A) and (B), the subject's sound side leg is shown (open squares), along with reference data from unimpaired walkers (standard error bars). For amputee participants 1 and 2 (Table I), the non-adaptive, mechanical knee produced a maximum flexion angle that increased with increasing speed, far exceeding 70 degrees at the fastest forward walking speed, whereas the user-adaptive knee gave a maximum flexion angle that always was less than 70 degrees and agreed well with the unimpaired, biological data. However, for amputee participants 3 and 4 (Table I), there was generally poor agreement between the unimpaired knee data and the data from *both* prosthetic knees. However, for these amputee participants, the maximum swing flexion angle was always below the 70 degree biological threshold even when using the mechanically passive knee prosthesis.

Discussion

In order for individuals suffering from trans-femoral amputation to walk in a variety of circumstances, an external knee prosthesis must provide stance control to limit buckling when weight is first applied to the prosthesis. In addition, a knee prosthesis must provide swing phase control so that biologically realistic leg dynamics emerge during swing. Unlike a biological leg, an external knee prosthesis, using only local mechanical sensing, must accomplish both stance and swing control without direct knowledge of its user's intent. Rather, such a device must infer whether its user desires stance or swing behavior and predict when future stance/swing transitions should occur. Such a device must also determine when dramatic changes occur in the environment, as for example, when an amputee decides to lift a suitcase or change to a heavier shoe.

Using commercially available prosthetic knee technology, a prosthetist must program knee damping levels until a knee is comfortable, moves naturally, and is safe (Popović and Sinkjaer, 2000). However, these adjustments typically are not guided by biological gait data; therefore, knee damping may not be set to ideal values, resulting in undesirable gait movements. Still further, knee damping levels may not adapt properly in response to environmental disturbances. In this study, a MR knee prosthesis is presented that automatically modulates knee damping values to match the amputee's gait requirements, accounting for variations in forward speed, gait style and body size. We find that the user-adaptive knee successfully controls early stance damping, enabling amputees to undergo biologically-realistic, early-stance knee flexion. Additionally, we find that the knee constrains the maximum swing flexion angle to an acceptable biological limit. The results of this study support the hypothesis that a user-adaptive control scheme and local mechanical sensing are all that is required for amputees to walk with an

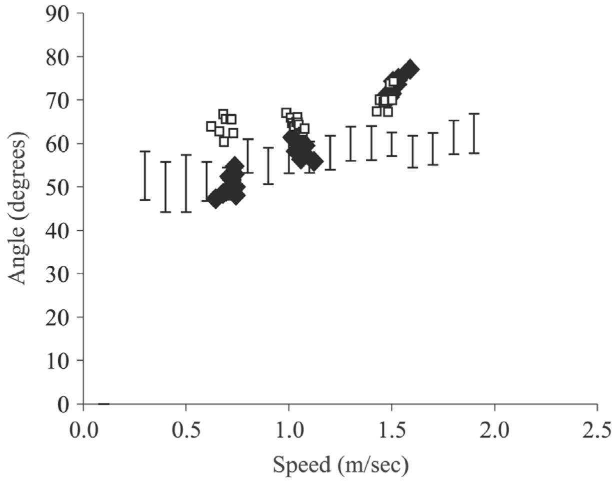
Industrial Robot: An International Journal
Volume 30 · Number 1 · 2003 · 42–55

increased level of biological realism compared to mechanically passive prosthetic systems.

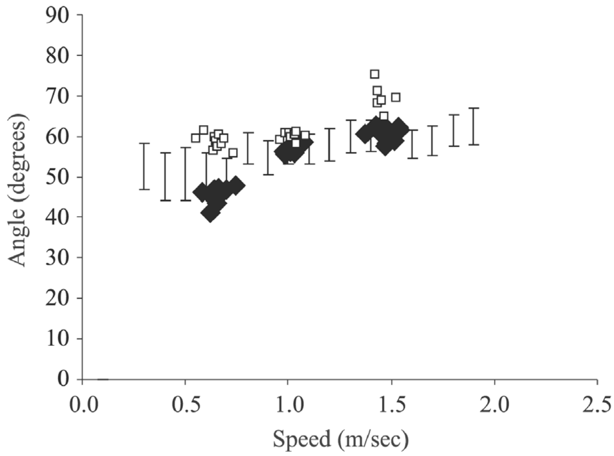
Dissipative knees and swing phase gait symmetry

Although the peak swing flexion angles measured from amputee participants 1 and 2 agreed well with the unimpaired, biological data (Figures 6 and 7), the data from amputee participants 3 and 4 did not. The peak swing flexion angles from these participants were generally much lower than the unimpaired, biological data even when using the non-adaptive passive knees (Figures 8 and 9). Since the electronic knee of this investigation is purely dissipative and cannot power movement, a peak swing flexion angle that falls below the 70 degree threshold cannot be increased by the system if the knee's active damping is already set to zero (electromagnet current = 0). Hence, for participants 3 and 4, the electronic knee controller set flexion damping equal to zero in an attempt to *increase* both the peak flexion angle and the level of gait symmetry between affected and unaffected sides. Only changes to the knee actuator design, not control system, would improve the biological realism and gait symmetry of participants 3 and 4. To increase the peak flexion angle for these individuals, the zero-current resistive torque

Figure 7 Maximum flexion angle during the swing phase versus walking speed for subject 2 (see Table I)



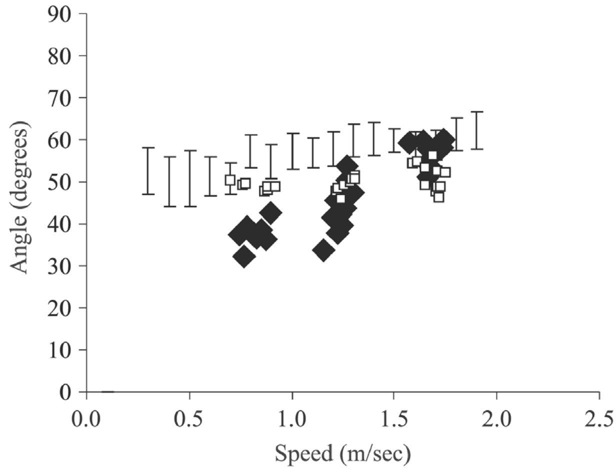
(A)



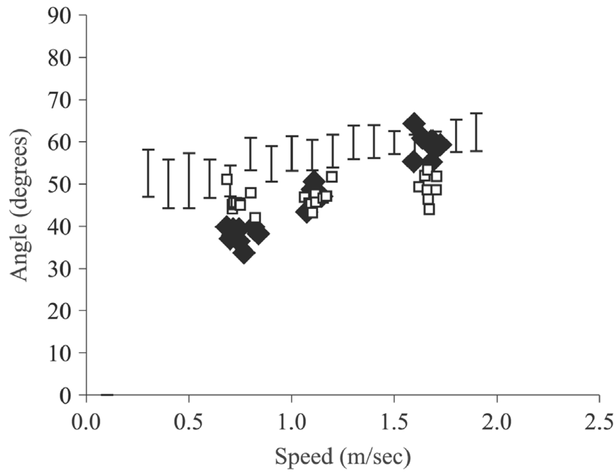
(B)

Note: Using the non-adaptive, mechanical knee (plot A, filled diamonds) and the adaptive knee (plot B, filled diamonds). In (A) and (B), the subject's sound side leg is shown (open squares), along with reference data from unimpaired walkers (standard error bars). In (A), the peak angle increases with increasing speed, exceeding seventy degrees, but in (B) the maximum flexion angle is less than seventy degrees and is in reasonable agreement with biological data

Figure 8 Maximum flexion angle during the swing phase versus walking speed for subject 3 (see Table I)



(A)

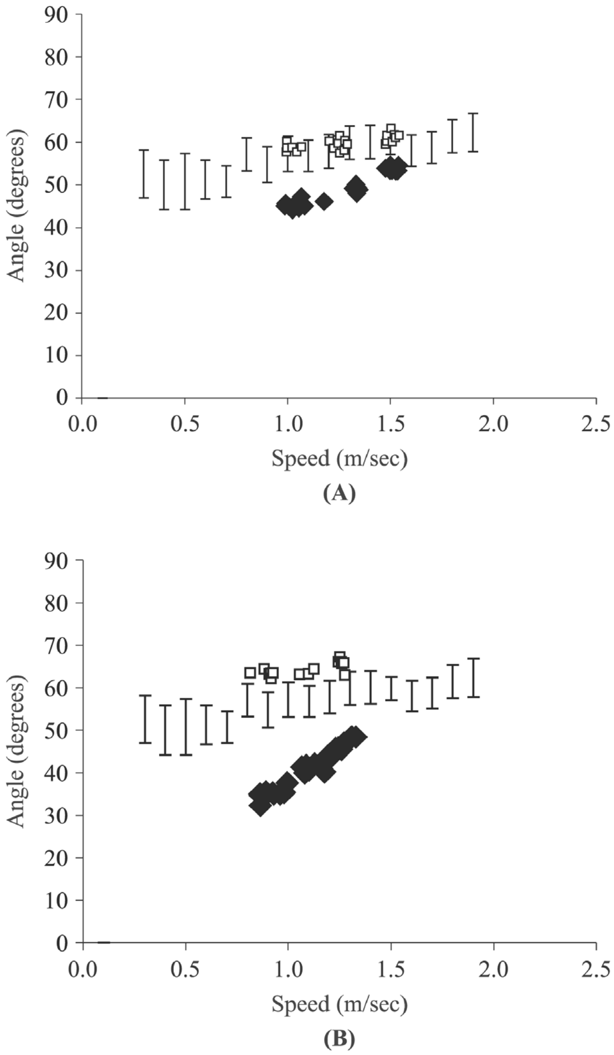


(B)

Note: Using the non-adaptive, mechanical knee (plot A, filled diamonds) and the adaptive knee (plot B, filled diamonds). As in Figures 6 and 7, the subject's sound side leg is shown (open squares), along with reference data from unimpaired walkers (standard error bars). In both (A) and (B), maximum flexion angle is low compared to the biological data except at the fastest walking speed

Industrial Robot: An International Journal
Volume 30 · Number 1 · 2003 · 42–55

Figure 9 Maximum flexion angle during the swing phase versus walking speed for subject 4 (see Table I)



Note: Using the non-adaptive, mechanical knee (plot A, filled diamonds) and the adaptive knee (plot B, filled diamonds). Here again, the subject's sound side leg is shown (open squares), along with reference data from unimpaired walkers (standard error bars). In both (A) and (B), maximum flexion angle is low compared to the biological data even at the fastest walking speed

of the actuator would have to be further decreased so that less of the leg's kinetic energy would be dissipated during late stance and early swing. However, the zero-current resistive torque of the electronic knee of this investigation is already low (0.5 Nm at physiologic knee angular rates). Thus, absolute gait symmetry during the swing phase may necessitate a knee actuator design that is truly active where both knee damping and power generation are modulated throughout a walking cycle.

Active knee prostheses

In their groundbreaking work, Popović and Schwirtlich (1988) developed an active trans-femoral prosthesis called the Belgrade Knee. With this system, both knee dissipation and mechanical power generation could be controlled throughout a walking step. Upon evaluating the clinical benefits of their prosthesis, they discovered dramatic improvements compared to conventional, mechanically-passive trans-femoral prostheses (Otto Bock 3P23 and Mauch SNS). Using a biomimetic control strategy, biological realism and gait symmetry were achieved. Additionally, the amputee participants could walk at a faster pace and with an improved metabolic economy. Although dramatic clinical advantages were achieved, the Belgrade knee was never commercialized because of problems with the mechanical system, including a limited battery-life. Acting autonomously, an amputee could only use the prosthesis for 3 h before the onboard battery had to be recharged (Popović and Kalanović, 1993). In contrast, variable-damper knee prostheses such as the user-adaptive system presented here, are more energy efficient, allowing amputees to use the prosthesis for an entire day before a battery recharge is necessary.

Future work

An area of future research of considerable importance is the development of improved power supplies and more efficient knee actuator designs where both joint dissipation and mechanical power generation can be effectively controlled in the context of a low-mass, high fatigue-life, commercially viable knee prosthesis. Another important area of research will be to combine local mechanical sensing about an external prosthetic knee with peripheral and/or central neural sensors positioned within the body. Neural prostheses such as the Bion (Loeb, 2001), combined with external biomimetic prosthetic systems, may offer important functional advantages to trans-femoral amputees. The fact that only local mechanical sensors were employed in the electronic knee of this investigation led to dramatic limitations in the systems ability to assess user intent. Such a prosthesis cannot determine whether a patient wishes to turn to the right or to the left, or whether an obstacle falls directly in the amputee's intended pathway. In the advancement of knee systems, we feel improvements in power supplies, knee actuation strategies and distributed sensory architectures are research areas of critical importance.

Concluding remarks

In this paper, we ask whether a computer-controlled, variable-damper electronic knee, employing only sensory information measured local to the knee axis, can automatically adapt knee damping values to match the amputee's gait requirements, accounting for variations in forward walking speed, user gait styles and body size. We show that a user-adaptive control scheme and local mechanical sensing are all that is required for amputees to walk with an increased level of biological realism compared to mechanically passive prosthetic systems. The user-adaptive knee successfully controlled early stance damping, enabling the amputee participants to undergo biologically-realistic, early-stance knee flexion. Additionally, the user-adaptive knee constrained the maximum swing flexion angle to an acceptable biological limit. It is our hope that this work will lead to further studies in prosthetic design resulting in an even wider range of locomotory performance advantages.

References

Aeyels, B., Peeraer, L., Sloten, V. and Van der Perre, G. (1992), "Development of an above-knee prosthesis equipped with a microcomputer-controlled knee joint: first test results", *PD Bioengineering Proceedings of Engineering Systems Design and Analysis Conference*, June 29-July 3.

Aeyels, B., Van Petegem, W., Sloten, V., Van der Perre, G. and Peeraer, L. (1995), "EMG-based finite state approach for a microcomputer-controlled above-knee prosthesis", *Annual International Conference of the IEEE Engineering in Medicine and Biology, Proceedings of IEEE Engineering in Medicine and Biology.*

Bar, A., Ishai, G., Meretsky, P. and Koren, Y. (1983), "Adaptive microcomputer control of an artificial knee in level walking", *J. Biomed. Eng.*, Vol. 5.

Darling, D., (1978), "Automatic damping profile optimization for computer controlled above-knee prostheses", Masters thesis Department of mechanical engineering, Massachusetts Institute of Technology.

Davis, R., Ounpuu, S., Tyburski, D. and Gage, J. (1991), "A gait analysis data collection and reduction technique", *Human Movement Science*, Vol. 10, pp. 575-87.

Deffenbaugh, B., Herr, H., Pratt, G. and Wittig, M. (2001), *Electronically Controlled Prosthetic Knee*, US Patent Pending.

Dietl, H. and Bargehr, H. (1997), "Der Einsatz von Elektronik bei Prothesen zur Versorgung der unteren Extremitat", *Med. Orth. Tech.*, Vol. 117, pp. 31-5.

Gard, A. (1999), "The influence of stance-phase knee flexion on the vertical displacement of the trunk during normal walking", *Archives of Physical Medicine and Rehabilitation*, Vol. 80.

Grimes, D. (1979), "An active multi-mode above-knee prosthesis controller", Doctoral thesis, Department of mechanical engineering, Massachusetts Institute of Technology.

Inman, V. (1981), *Human Walking*, Waverly Press.

James, K., Stein, R., Rolf, R. and Tepavac, D. (1990), "Active suspension above-knee prosthesis", *Goh JC 6th Int. Conf. Biomech. Eng.*, pp. 317-20.

Ju, M., Yi, S., Tsuei, Y. and Chou, Y. (1995), "Fuzzy control of electrohydraulic [sic] above knee prosthesis", *JSME International Journal, Series C: Dynamics, Control, Robotics, Design and Manufacturing*, Tokyo, Japan, Vol. 38 No. 1, pp. 78-86.

Kadaba, M., Ramakrishnan, K. and Wootten, M. (1990), "Measurement of lower extremity kinematics during level walking", *Journal of Orthopaedic Research*, Vol. 8, pp. 383-92.

Kastner, J., Nimmervoll, R., Kristen, H. and Wagner, P. (1998), "A comparative gait analysis of the C-Leg, the 3R45 and the 3R80 prosthetic knee joints", http://www.healthcare.ottobock.com

Kautz, T. and Seireg, A. (1980), "Feasibility study of a computer controlled hydraulic above-knee prosthetic limb", *International Conference on Med. Devices and Sports Equip., Century 2 - Emerging Technology Congress*, San Francisco.

Loeb, J. (2001), "Neural prosthetics", in Arbib, M.A. (Ed.), *The Handbook of Brain Theory and Neural Networks*, 2nd ed., MIT Press, Cambridge, MA.

Myers, D. and Moskowitz, G. (1981), "Myoelectric pattern recognition for use in the volitional control of above-knee prosthesis", *IEEE Trans. Syst. Man Cybern.*, Vol. SMC-11 No. 4, pp. 296-302.

Myers, D. and Moskowitz, G. (1983), "Active EMG-controlled A/K prosthesis. Control aspects of prosthetics and orthotics", *Proceedings of the IFAC Symposium*, Columbus, OH.

Peeraer, L., Aeyels, B. and Van der Perre, G. (1989), "Development of EMG-based mode and intent recognition algorithms for a computer-controlled above-knee prosthesis", *Journal of Biomedical Engineering, 29th Annual Meeting of the Biological Engineering Society: Engineering for Health*, Vol. 12 No. 3, pp. 178-82.

Peeraer, L., Tilley, K. and Van der Perre, G. (1988), "Computer-controlled knee prosthesis: a preliminary report", *Advances in Rehabilitation Technology*, 13-15 April, Vol. 13 No. 1.

Popović, D. and Kalanović, V. (1993), "Output space tracking control for above-knee prosthesis", *IEEE Transactions on Biomedical Engineering*, Vol. 40 No. 6, pp. 549-57.

Popović, D. and Schwirtlich, L. (1988), "Belgrade active A/K prosthesis", in de Vries, J. (Ed.), *Electrophysiological Kinesiology*, Interm. Congress Ser. No. 804, Excerpta Medica, Amsterdam, The Netherlands, pp. 337-43.

Popović, D. and Sinkjaer, T. (2000), *Control of Movement for the Physically Disabled*, Springer-Verlag, London.

Popović, D., Oguztoreli, M. and Stein, R. (1991), "Optimal control for the active above-knee prosthesis", *Annals of Biomedical Engineering*, Vol. 19 No. 2, pp. 131-50.

Ramakrishnan, H., Kadaba., M. and Wootten, M. (1987), "Lower extremity joint moments and ground reaction torque in adult gait", in Stein, J.L. (Ed.), *Biomechanics of Normal and Prosthetic Gait, ASME Winter Annual Meeting*, BED-Vol. 4, pp. 87-92.

Triolo, R. and Moskowitz, G. (1982), "Autoregressive EMG analysis and prosthetic control", *Proceedings of the 35th Annual Conference on Engineering in Medicine and Biology*, Philadelphia, PA.

Wilkenfeld, A. An auto-adaptive external knee prosthesis Doctoral Thesis Department of mechanical engineering, Massachusetts Institute of Technology.

Further reading

Goldfarb, M. (1992), "Control of a self-contained microcomputer-controlled above-knee prosthesis" Masters Thesis, Department of mechanical engineering, Massachusetts Institute of Technology.

Herr, H., Wilkenfeld, A. and Olaf, B. (2001), *Speed-Adaptive and Patient-Adaptive Prosthetic Knee*, US Patent Pending.

Qi, D. (1986), "A speed adaptive control algorithm for the self-contained A/K prosthesis" Masters thesis, Department of mechanical engineering, Massachusetts Institute of Technology.