ADAPTIVE MICROCOMPUTER CONTROL OF AN ARTIFICIAL KNEE IN LEVEL WALKING

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ABSTRACT

An experimental microcomputer controlled prosthesis is discussed, where knee damping is governed by kinematic signals, picked up from both legs. The control algorithm is based on division of the walking cycle into ten stages; each stage characterized by an appropriate damping level which governs the restraining moment at the knee, as functionally required. Stage identification is carried out by the kinematic inputs. The experimental prosthesis was tested on one amputee, the first author of this paper. The control system was found to be more advantageous than a conventional valve system in significantly reducing hip muscular effort on the prosthetic side. Gait symmetry is also improved.

Keywords: Prosthetic devices; knee joint, gait analysis, microcomputer, control

INTRODUCTION

The large difference in gait performance between below-knee and above-knee amputees reflects the importance of the knee in locomotion control and illustrates the shortcomings in performance of existing artificial knee mechanisms. Efforts are continuously being made to develop new knee mechanisms to alleviate amputee discomfort, lower energy consumption and instantaneous muscular effort, reduce loads transferred to the sound leg and to the vertebral column, and improve stability and appearance in walking.

Following the loss of the physiological knee, the hip muscles must control the new knee as well as continuing to control thigh-pelvis angular motion. Within the walking cycle, there exists a conflict between knee and hip control requirements. Regarding hip moments, the choice of moment is by the dominant requirement which leaves the other joint unsatisfied and causing a deficient gait. Knee mechanisms are designed to complement hip muscles in knee control, utilizing ground reactions and shank inertia to adjust moment transfer about the knee axis. The design of one category of artificial knees, is based exclusively on mechanisms with mechanical properties (such as friction, spring and damping coefficients) which remain constant during the walking cycle or depend on knee flexion-extension kinematics. A common method of achieving knee stability during stance, in prostheses of this category, is based on locating the knee axis more posteriorly than the physiological knee, thus utilizing ground reaction to assist the hip muscles in locking the knee. An associated disadvantage is the large hip flexion moment required to bend the knee near the end of the stance phase. Polycentric knees are used to reduce the required hip moments for both stabilizing and bending the knee during stance, by controlling instantaneous knee centre location as a function of knee angle[1-3].

Swing control is usually carried out in AK prostheses of this category by dissipating energy, using constant friction or a hydraulic damping, and by storing energy in a spring during knee flexion and releasing it during extension. The use of these three mechanical principles, produce knee moments according to knee angular kinematics, which satisfy swing requirements, such as ground clearance and cadence control, do so for a so-called 'standard gait' in a limited range of walking speed. Another category of AK prostheses includes knee mechanisms with mechanical properties which are controlled by signals arriving from sources outside the knee itself, in addition to those which depend on knee angular kinematics. For example, stance phase stability in the Lin*, Regnel† and Mauch's[4] SNS hydraulic knee is achieved by locking the knee during stance. The trigger signal for unlocking in the Lin and Regnel prostheses occurs near the end of stance phase when dorsi flexion at the ankle joint reaches a predetermined level. In the SNS prosthesis knee flexion damping is significantly reduced when the hyperextension moment about the knee reaches a preset value.

The feasibility of using EMG signals picked up from stump muscles, for knee control, has been studied by several investigators. Horn[5] and Sexena[6] used EMG signals for the control of knee stability during stance. Dyck[7] used an EMG controlled hydraulic knee with four damping levels, for both swing and stance control.

EMG for knee control seems attractive. It may provide voluntary conscious control of knee function, utilizing EMG patterns inherent in the stump during gait. The main disadvantages of EMG knee control are the required concentration and the intricacy of reliable EMG signal selection.

The introduction of a computer into the knee control system allows emulation of proposed knee

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designs and control concepts without having to develop hardware. Furthermore, the use of a microcomputer in future commercial prostheses will make possible an increase in the number of feedback signals. It will also be practical to apply control strategies which adapt knee moments to the varying functional gait requirements. Radcliffe[8] developed an AK prostheses which utilizes a minicomputer to govern a servo valve which regulates a hydraulic knee damper. Knee locking was triggered by the cessation of extension angular velocity and unlocking was voluntarily achieved by a hip flexion moment greater than a preset value. The use of hip flexion moment as a control signal for knee stability, allows conscious timing control of knee unlocking in unexpected situations during gait, or in intricate activities such as descending stairs, as well as automatic triggering of knee unlocking during 'standard gait'. Flowers[9] developed two computer controlled knee prostheses. In one system, swing and stance control were effected via an electro-hydraulic actuator. The second utilized a magnetic practical brake. These systems were used to evaluate the following control concepts: (a) The control of knee damping via myoelectric activity of a residual muscle of the stump, (b) application of predetermined knee moment characteristics as function of knee angular kinematics, and (c) duplication of the angular position pattern of normal knee during stance phase[10]. The two latter control concepts could have been suitable for purely cyclic gait; nevertheless, it seems that as walking is not strictly repetitive, knee control should be related, in addition to knee angle, to other variables reflecting instantaneous required function of the prosthetic leg as a whole.

Drawbacks still exist in the reviewed methods of knee control mainly because of the difficulty in adapting knee moments to the instantaneous functional requirement of the prosthetic leg during the whole walking cycle. The problems are: 1) difficulty in identifying control signals which characterize an instantaneous kinematic state of the locomotor system, 2) dynamically altering the mechanical properties of the artificial knee.

The aims of the present work were:

(a) to define the major functional requirements of the prosthetic knee during the whole walking cycle;

(b) to identify measurable kinematic variables which may be related to, and thus characterize, the instantaneous functional requirements;

(c) to develop an experimental prosthesis which employs these variables as control signals to govern knee moment via a microcomputer.

(d) to evaluate the developed system vis à vis conventional prostheses.

THE EXPERIMENTAL PROSTHESIS

A control system was designed in this particular case to improve knee performance in level walking, fulfilling both stance phase requirements, i.e. knee stability, controlled rate of knee flexion, free forward swinging of the stump as well as swing requirements, i.e. ground clearance and swing duration control. These requirements should be satisfied to minimize both exertion of hip muscles and 'compensatory movements' of the trunk and sound leg.

The assembled prosthesis is shown in *Figure 1* and a block diagram describing the operation of the control system is depicted in *Figure 2*. Knee flexion - extension - resisting moment is supplied by a microcomputer controlled hydraulic damper, according to kinematic signals, picked up from both legs. The knee mechanism is based on a commercially available four-bar prosthesis (OHC, a product of USMC) comprising a Dyna-Plex damper with an extension spring. The damper's needle valve has been replaced by a specially developed digital valve,

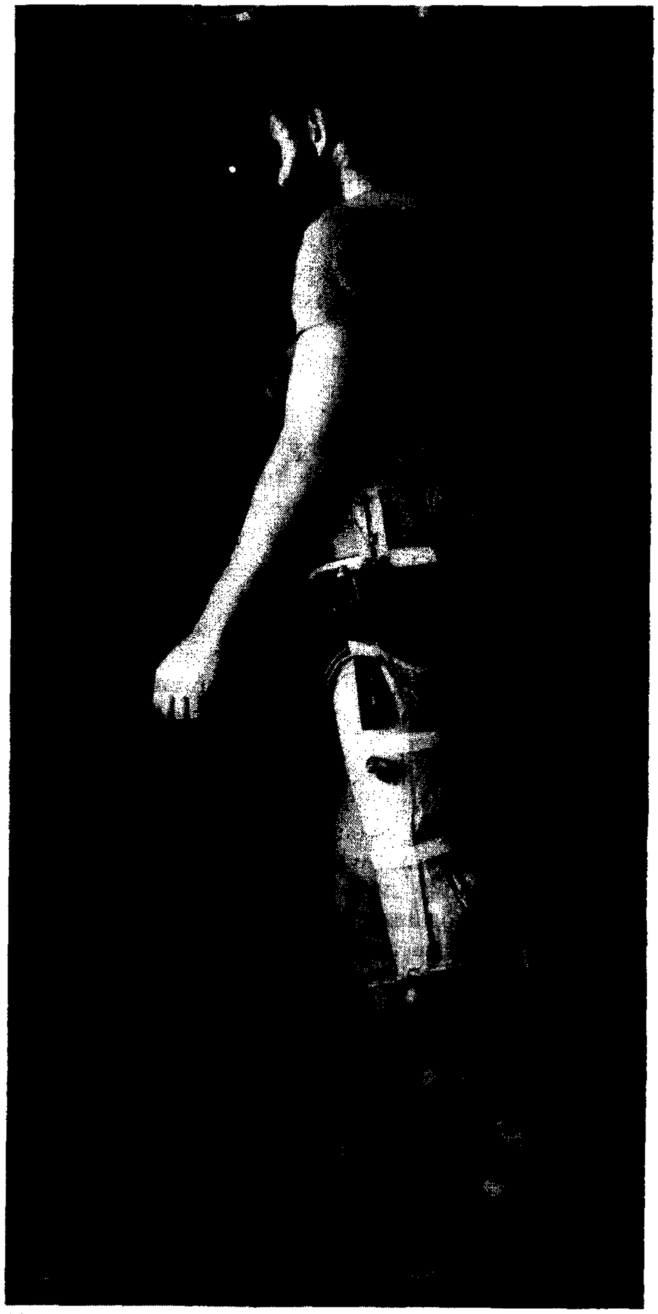

Figure 1 An amputee wearing the experimental prosthesis.

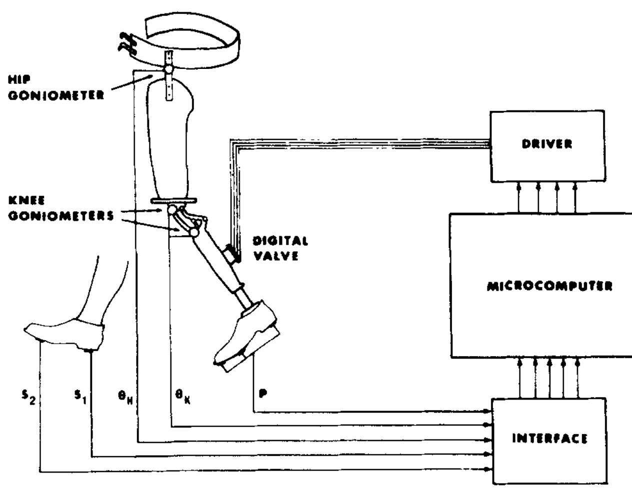

Figure 2 Control block diagram.

having four normally closed orifices, each opened by a solenoid. A computer algorithm selects which of the sixteen possible valve states is to be activated by suitably addressing the electronic solenoid drivers. The choice of damping level is made according to the kinematic state of both legs, as determined by five control signals sampled by the microcomputer at a 50 Hz rate. The control signals are: (1) The angle between the socket and pelvis (θ_H), measured by a potentiometer which is attached to the socket and to a pelvic belt which measures nothing but hip flexion extension angle. (2) Knee angle (θ_K), is measured by potentiometers, attached to two of the four joints that comprise the knee mechanism. (3) Prosthetic foot contact with the ground (P) is measured by means of three shoe switches installed at heel, mid-sole and fore-foot. Closure of one or more of the switches indicates prosthetic foot contact. Two additional switches are used as separate control signals to indicate: (4) heel contact (S_1) and (5) fore-foot contact (S_2) of the sound leg.

The control algorithm is based on division of the walking cycle into ten stages, each characterized by a predetermined damping level. The stages are defined in terms of the analogue variables θ_H, θ_K and $\dot{\theta}_K$ (rate of change of knee angle), and the digital variables P, S_1 and S_2 ('1' contact, '0' no contact) as summarized in *Table 1*.

The basic concept behind stance control in the prosthesis is that knee locking is to be achieved independently of the magnitude and orientation of ground reaction force while knee bending is to be activated spontaneously by ground reaction with minimal hip flexion moment exertion. This concept is applied by locating the knee axis anteriorly to the usual location so that knee flexion moment caused by ground reaction starts earlier than in commonly used prostheses. Knee locking and unlocking are triggered by the sampled kinematic signals, and knee flexion is initiated following the appearance of a knee-unlock signal, when ground reaction applies a flexing moment about the knee.

The rate of knee flexion during stance is controlled

Table 1 Stage identification by algorithm according to measured kinematic state of both legs. (P – prosth. foot contact; S_1 – sound leg heel contact; S_2 – sound leg fore-foot contact; θ_K, $\dot{\theta}_K$ – prosth. knee angle & angular velocity; θ_H – prosth. stump/pelvis angle).

Stage	P	S_1	S_2	θ_K	$\dot{\theta}_K$	θ_H
1	1	0	0	/	/	$> \theta_{H1}$
2	1	0	0	/	/	$\leqslant \theta_{H1}$
3	1	1	0	/	/	$\leqslant \theta_{H1}$
4	1	/	1	/	/	$\leqslant \theta_{H1}$
5	0	/	/	$\leqslant \theta_{K1}$	$\geqslant 0$	/
6	0	/	/	$> \theta_{K1}$	$\geqslant 0$	/
7	0	/	/	/	< 0	$\leqslant \theta_{H2}$
8	0	/	/	$\geqslant \theta_{K2}$	< 0	$> \theta_{H2}$
9	0	/	/	$\theta_{K3} < \theta_K < \theta_{K2}$	< 0	/
10	0	/	/	$< \theta_{K3}$	/	/

The values for the specific subject were: $\theta_{H1} = 7°$; $\theta_{K1} = 60°$; $\theta_{K2} = 5°$; $\theta_{K3} = 1°$

by applying appropriate damping at the mechanism. In choosing damping levels during stance phase, the following considerations have to be accounted for: low damping is desired to enable early knee bending, reducing the peak value of hip extension angle at the end of stance phase, shortening the time required to swing the thigh forward at later stages and lowering the trajectory of the body centre of mass. Conversely too little damping may cause the amputee to feel unstable and cause excessive loading rate on the sound leg.

Stance and swing phases on the prosthetic leg have been divided into four and six stages, respectively. Stage 1 starts when heel contact of the prosthetic leg is detected ($P = 1$) and ends when the hip angle θ_H reaches a preset negative value ($\theta_H = -\theta_{H1}$) which corresponds to knee unlocking. During this stage, knee stability is achieved by closing all solenoid valve orifices which hydraulically lock the knee.

The next three stages are concerned with prosthesis stance to swing phase transition control. During these stages the knee must bend, while the prosthesis still bears all, or part, of body weight.

Stage 2 starts when $\theta_H = -\theta_{H1}$ and ends with the detection of sound leg heel contact on the sound leg ($S_1 = 1$). During this stage the knee is unlocked. High damping is required as total body weight is supported by the prosthetic leg. Knee bending is necessary towards the end of single support to prevent body 'pivot' over the prosthetic fore-foot.

Knee axis location was adjusted to give small flexion moment about the knee at 'normal' standing due to ground reaction. The value of hip extension angle θ_{H1}, which triggered knee unlocking, was set outside the range of hip angles present in standing, thus ensuring knee locking at standing while permitting early knee bending in walking.

Stage 3 starts at the initiation of double support as detected by heel contact of the sound leg ($S_1 = 1$).

It ends when the fore-foot of the sound leg hits the ground ($S_2 = 1$). At this stage, knee damping is reduced as compared to stage 2, because the sound leg starts to bear load and shares in stability control.

Stage 4, the last of the stance phases, starts at 'foot-flat' of the sound leg ($S_2 = 1$) and ends when contact between the prosthetic foot and the ground ceases ($P = 0$). Damping is further reduced at this stage, to avoid dragging the top of the foot during transition to the swing phase.

The main considerations in establishing knee control strategy for the swing phase were concerned with (a) reduction of swing period, (b) achievement of ground clearance without investing excessive compensatory body movements and (c) reduction of hip muscular involvement in flexing and extending the knee.

During the first two stages of the swing phase, i.e. stages 5 and 6, knee flexion angle reaches its maximal level, i.e. large enough to give sufficient ground clearance at later stages, and small enough to shorten swing period. Stage 5 starts at the initiation of swing phase ($P = 0$) and ends when knee angle reaches a preset level ($\theta_K = \theta_{K1}$) quite close to maximum required flexion. To ensure sufficient flexion in minimal time, the damping during this stage, is at a minimum. Stage 6 is present in the control routine only if $\theta_K = \theta_{K1}$ is reached, at non zero angular velocity ($\dot{\theta}_K > 0$), and ends at full flexion ($\dot{\theta}_K = 0$); during this stage high damping is applied to limit knee flexion.

Stage 7 starts at the beginning of knee extension ($\dot{\theta}_K < 0$) and ends when hip flexion angle reaches a preset positive value ($\theta_H = \theta_{H2}$), which means the foot has already passed its lowest position and cleared the ground. Here high damping is applied to delay knee extension. To guarantee ground clearance, the foot was located posteriorly from normal location; possible because knee stability during stance does not depend on the location of ground reaction relative to the leg.

When $\theta_H = \theta_{H2}$ is reached, stage 8 starts. Now the knee is extended with minimal damping. The energy stored in the extension spring is released, assisting in swinging the shank forward. Due to the low damping at both flexion and extension of the knee (stages 5 and 8), and due to the delay caused to knee extension at stage 7 to ensure ground clearance, the amount of elastic energy which may be utilized for knee extension, is much larger in the present system as compared to systems incorporating a single valve setting. The extra energy storage was achieved by high prestressing of the extension spring, compensating for the delay presented by stage 7, to shorten the swing phase period.

Stage 9 covers the last five degrees of knee extension ($0^\circ < \theta_K < 5^\circ$). The damping in this stage is increased to attenuate the impact occurring at full knee extension. Stage 10, the last control stage, starts immediately when full knee extension is reached. The knee is being hydraulically locked to prepare it for load carrying at heel contact (stage 1) and to avoid the need for exertion of a hip extension moment as commonly used at this stage by AK amputees. In cases where full extension is not reached at the end of swing phase, the knee locks at heel contact ($P = 1$, stage 1).

EXPERIMENTAL RESULTS

Experiments were conducted to optimize the constructed system's control parameters and to evaluate the prosthesis performance in walking. Gait trials were performed on the same through-knee amputee, the first author of this paper, whose normal prosthesis was basically the same as the experimental one, except that it uses a manually adjusted pre-set orifice to control hydraulic damping. The trials were confined to the laboratory because of umbilical links to the power supplies and to the micro computer (Sintel 8085). The parameters to be optimized were: (1) threshold levels of knee and hip angles and knee angular velocity, which define transitions between control stages. (2) Foot switch locations on the sole of the shoes, which determine timing of stage transitions. (3) Damping level at each control stage. (4) Antero-posterior location of knee axis, determining the time when knee bending starts. (5) Location of the foot relative to the shank ensuring ground clearance and affecting initiation of knee bending. (6) Coefficient and amount of pre-stress of the extension spring. Changes in adjustment of control parameters were followed by training periods. These were found, in many cases, to be longer than accommodation periods for standard alignment sessions, due to differences in both control principle and amputee's sensation differences between the normal and the experimental prosthesis. The main phenomenon which required long accommodations was early knee bending during stance (stage 2). Decisions regarding changes in adjustment of control parameters were based mainly on the amputee's subjective evaluation of the situation. To assist in the interpretation of the control system's operation, the control signals θ_K, θ_H, P, S_1 and S_2, together with the valve command signals were sampled by a PDP 11/55 minicomputer at a rate of 50 Hz; the sampled data was displayed on a graphic terminal in the form shown in *Figure 3*. In the early stages of the experimental work, the ratio between orifice areas in the solenoidic valve was 1:2:4:8, giving sixteen damping levels (including complete locking.) The smallest orifice was 0.6 mm in diameter. System performance was satisfactory at low damping levels but better resolution was necessary at higher levels. Several combinations of orifice sizes were tested. The most satisfactory results occurred for orifice areas ratio of 1:1.36:2.77:9, the smallest orifice remaining constant as before. This combination was also satisfactory in view of the control strategy. It was found that a change in damping levels requiring some apertures to close while others were opening caused a non-monotonic change in damping delay because of opening-closing differences. Such transitions caused discomfort to the amputee. The ultimate ratio between orifice areas led to monotonic transitions for the specific prosthesis control strategy. The performance of the system at walking speeds of 2.0, 3.6 and 5.1 km/h, is depicted in *Figure 3*, 3.6 km/h being 'free walking' velocity. The figure represents optimal settings of the control parameters each arrived after repeated walking trials at the specific speed. The valve state curves represent the total area of valve openings. No changes of control parameters other than damping were made at a transition from one speed to another. It is evident that differences in optimal valve states exist towards the end of stance phase and at about mid-swing, for different walking speeds. In states 2, 3 and 4, lower damping is required as walking speed is increased. This is explained by the larger knee-angular-velocity required as speed increases. Another phenomenon observed from *Figure 3* is that the time interval between Foot-Flat (FF) on the sound leg and Toe-Off (TO) of the prosthesis shortens as walking speed increases, which shortens stage 4 to zero at 5.1 km/h. It should be noted that inclusion of foot switch signals from the

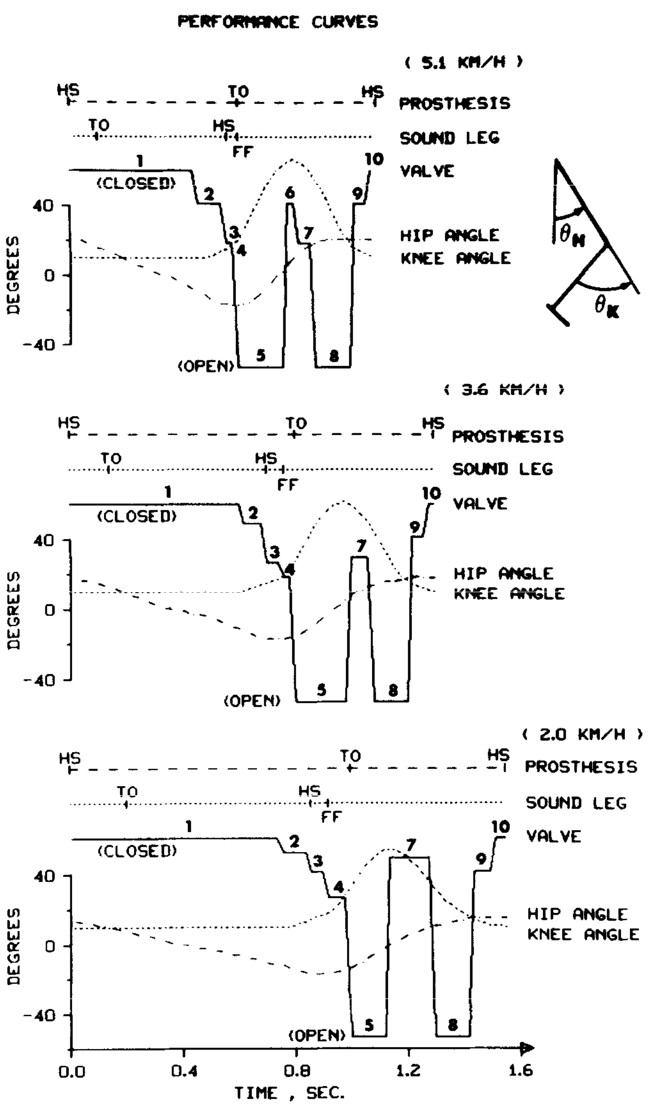

Figure 3 Valve state as related to kinematic inputs during a single walking cycle, at three walking speeds.

sound leg (S_1 and S_2) as control signals, significantly improved walking performance. Initially, such switches were not included, and high damping was maintained until nearly the end of prosthetic stance phase, with consequent difficulties in knee bending. Direct information of sound leg's heel-contact and foot-flat, permitted a decrease in knee damping without compromising walking stability.

During swing phase, stage 6 was found to occur exclusively at high walking speed. At lower speeds, knee flexion never reached the threshold level θ_{K1}. It was not necessary to reduce θ_{K1} at lower speeds, because resistance of the extension spring was sufficient to limit knee flexion. In stage 7, the damping level had to be reduced with increasing walking speed, due to the larger knee flexion angle obtained at higher speeds. The resulting duration of stage 7 is shorter with increased speed since hip flexion causes the threshold level of θ_{H2} to be reached more quickly.

The newly developed control system has proven to be significantly advantageous when compared to the single valve system of a normal prosthesis. The two most appreciated features of the controlled prosthesis were: (1) lower effort required from prosthetic side hip muscles for knee stabilization and bending initiation. (2) the sensation of thigh motion equality in both legs.

CONCLUSIONS

The aim of the presented project was to evaluate the concept of "adaptive knee control" in level walking. Emphasis was given to the choice of control signals and algorithm and to the practical aspects concerned with the manufacturing of a self contained prosthesis for general use. For example, the use of multiple goniometers and foot switches does not lend itself to general prosthesis applications. Experiments were confined to the laboratory and the amputee was connected by umbilical links to the power supplies and to the microcomputer.

At present, the system has been tested on just one amputee, the first author of this paper. Further experiments will be carried out on several amputees, in order to examine the universality of the algorithm and the function of personal biases. It should be noted that selection of the system's parameters is required for any individual amputee.

The following conclusions were derived from the level walking trials:

1. Integrating measured kinematic information for walking-stage identification, makes it possible to precisely and reliably adapt the level of knee damping to the instantaneous functional requirement for the knee. To achieve reliability without compromising functional quality, the control strategy must rely on both knee kinematics and 'external' signals arriving from other sources.

2. Dependence of knee control and function on the measured kinematic state of both legs, reduces involvement of hip muscles in knee control, bringing about lower hip muscles effort, and improved symmetry of thigh motion of both legs.

3. Damping level characteristic must vary in accordance with walking speed.

4. The advantages of the digital solenoid valve were: (a) low time response and (b) simple control circuitry. Nevertheless, it should be replaced by an analogue valve, because of difficulties in achieving good resolution at high damping and in obtaining qualitatively "smooth" transitions between damping levels.

   In order to bring the system to a state which permits experiments outside the laboratory and, in the long run, as a system for general use, further developments should be carried out as follows: a. The operating system should be reduced in size, weight and energy consumption, b. Both microcomputer and electrical power supply should be installed within the prosthesis, c. The kinematic sensors should be improved both in structure and linkage to the control system, and d. The control algorithm should be expanded to include activities other than level walking, and to automatically adapt to variations in walking speed and to different activities.

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