A VOLUNTARILY CONTROLLED ELECTROHYDRAULIC ABOVE-KNEE PROSTHESIS

W.R. Dyck, M.Sc. [a]
Biomedical Engineer, Department of National Defence
Ralston, Alberta, Canada

S. Onyshko, Ph. D.
Associate Professor, Department of Electrical Engineering
University of Manitoba, Winnipeg, Manitoba, Canada R3T 2N2

D.A. Hobson, B.Sc. [b]
Technical Director, Rehabilitation Engineering Program
Department of Orthopaedics, University of Tennessee
858 Madison Ave., Memphis, Tenn. 37163

D.A. Winter, PH. D. [b]
Associate Professor, Department of Kinesiology
University of Waterloo, Waterloo, Ontario, Canada

A.O. Quanbury, M.Sc.
Assistant Director of Biomedical Engineering Research
Shriners Hospital, 633 Wellington Crescent
Winnipeg, Manitoba, Canada R3M OA8

ABSTRACT

Most above-knee amputees to date are using prostheses employing either constant friction or some type of programed hydraulic damping which the wearer has no control over and which limits his gait speeds.

[a] Formerly with the University of Manitoba, Winnipeg, Canada.
[b] Formerly with the Shriners Hospital, Winnipeg, Canada.

A new system was designed and tested in which voluntary control of a lower-limb prosthesis is derived from the EMG signals of residual thigh muscles in the stump. These signals, after suitable conditioning, open or close solenoid valves, which form a closed hydraulic loop around the damping cylinder in the knee joint. Thus, the amputee is able to voluntarily vary the resistance to knee flexion, from free swing to full lock, by operating valves controlling the resistance to flow around a hydraulic cylinder. The main advantages of this system are a variable and more aesthetic gait, stability over uneven terrain, and because the lock prevents only knee flexion, the amputee can rise on his prosthesis and so use a passive appendage as an active element of his skeleton.

The results of this project demonstrate that the concept of an EMG voluntarily controlled hydraulic prosthesis is viable; however, continuing effort is required to make this system lighter, more compact, and cosmetically acceptable.

INTRODUCTION

The design and development of an improved prosthesis for the above-knee amputee constitutes a challenging problem for the engineering designer. The prosthesis worn by an amputee is a specialized mechanism which becomes part of his daily activities and which must provide unfailing performance under many different conditions. Its design must therefore be based on both biological and engineering principles (1).

The advantages of the commonly used single-axis above-knee prosthesis with mechanical friction knee control are simplicity, reliability, and commercial availability at reasonable cost. However, its main disadvantages are related to the quality of swing-phase control, the need for a fully extended knee for stability during the weight-bearing, and the dependence of gait speed on the mass and geometry of the prosthesis.

Design development effort over the past few decades has attempted to solve the cadence limitation problem through the introduction of fluid damping systems, which create variable resistances about the knee joint. The prime advantage of the fluid control approach is related to the fundamental property of fluids. That is, if a fluid damper is located across the knee joint and is driven by the dynamic forces of the walking amputee, the system can be arranged so that the resistance created by the controlled flow of the fluid will vary with the angular velocity of the swinging shank. In addition, this resistance can be made to depend on the direction of fluid flow.

Pneumatic dampers have received considerable acceptance, primarily due to their simplicity. However, it is generally recognized that the characteristics provided by the controlled flow of a hydraulic fluid more closely approximate the forces required to yield the necessary damping resistance about the knee joint (2). The most successful results with hydraulic systems have been achieved in the Mauch Swing-N-Stance Knee (3). This system provides "programed" control of both swing and stance phases, all within one hydraulic knee unit.

The limitation of the Mauch system is that the knee control is only under partial control of the amputee, who must pass through certain knee activities in a predetermined pattern in order to achieve full control of the knee characteristics. The purpose of this research is to move forward and to develop a knee-control system which would be under the complete voluntary control of the above-knee amputee.

Research has begun on hydraulic dampers whose resistance can be voluntarily controlled by the wearer using either myoelectric or various mechanical signals (4). A recent development in above-knee prostheses by G. W. Horn (5), was an EMG-controlled flexion lock. Two advantages of this system are that a wearer can bear weight on a slightly bent knee allowing for a more normal appearing gait, and that a wearer can manage to climb stairs on his prosthesis in a normal way by not being restricted to taking one step at a time. This design does not, however, allow the amputee to vary the resistance in order to accommodate different gait speeds.

To accomplish this goal, the authors have developed a prototype system for a voluntarily controlled variable resistance hydraulic knee unit for an above-knee prosthesis, shown in Figure 1. A hydraulic damping cylinder is placed across the knee joint, and across this cylinder are a set of solenoid bypass control valves. The solenoids on these valves are then connected to an EMG control unit, which takes raw EMG signals and converts them into a signal which controls the energizing of the correct combination of valves. In this manner, an amputee will be able to voluntarily control the resistance to flexion and extension of the prosthesis from free swing to full lock.

DESIGN OF HYDRAULIC SYSTEM

An infinitely variable damper, proportionally controlled by the EMG activity of the amputee, was recognized as being the ideal system. However, this system presented technical problems beyond the scope of the project, and so a system with discrete damper control was considered as an acceptable compromise.

To determine the required characteristics of the hydraulic system a geometric model of the damping cylinder in the knee unit was established (Fig. 2). This model was used to convert known locomotion data of normal subjects to useful design information for the hydraulic system. This information included pressure rating of each component, rates of flow, resistance to flow versus percent walking cycle, and the number of bypass values and their resistance.

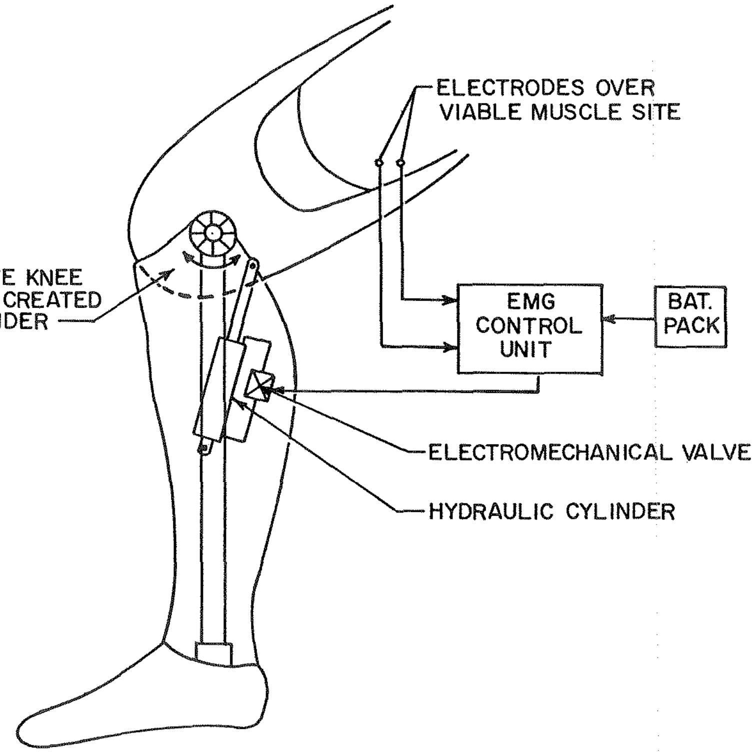

FIGURE 1.—Schematic of proposed voluntary controlled electrohydraulic prosthesis.

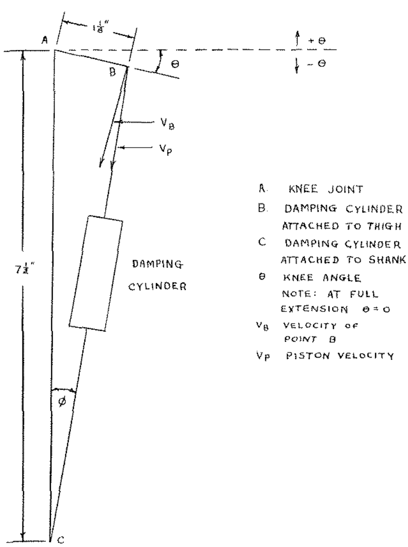

FIGURE 2.—Geometric scheme of hydraulic damping cylinder in knee unit.

The required force on the damping cylinder throughout the walking cycle was found by calculating the force needed in this cylinder with the given geometry of the cylinder linkage to produce moments about the knee corresponding to those measured in locomotion studies of normal subjects (Fig. 3). This knee moment data is for normal subjects, since such data was not available for amputees. From the time derivative of knee angle versus percent walking cycle the angular velocity of the knee was obtained, which permitted the determination of piston velocity versus walking cycle. This piston velocity was used to determine the flow rates in the hydraulic system after establishing that the cross-sectional area of the damping cylinder be 1 in^2, which is the largest that could be fitted into the knee unit conveniently.

The number of bypass valves and their flow resistances depends largely on the desired force in the damping cylinder during the walking cycle. A valve's resistance to flow is characterized by its C_V factor, which is defined as the flow of 60 deg. F. water (gal./min.) through the valve with a 1 p.s.i. pressure drop across it. The desired C_V needed across the damping cylinder was calculated and plotted, and an attempt was made to fit a valve system to this plot in order to determine the number and the type of valves needed.

Figure 4 shows the resultant C_V versus percent walking cycle plot, and also shows the fit to this plot which was accomplished by using two solenoid bypass valves and one check valve all in parallel with the damping cylinder. The two solenoid valves V1 and V2 have C_V factors of 0.02 and 0.1, respectively. The check valve, which permits flow only during extension, has a C_V factor of 0.08, sufficient to permit the high flow rate required during free swing and to avoid a lock in extension which was deemed undesirable. Because of the check valve the resistance to extension during weight-bearing (15–40 percent) is lower than necessary. However, it was felt more desirable to have resistance that is too low rather than too high, since with sufficient training the amputee should be able to control the terminal impact with his stump action.

BENCH TESTING OF HYDRAULIC SYSTEM

The objectives of bench testing the hydraulic system were: 1. to confirm that the system actually met the design criteria of force versus percent walking cycle, 2. to observe the hydraulic system operation under varied (simulated) walking speeds, and 3. to determine the effect of altering the switching sequence of the bypass valves.

A test jig was designed and built to simulate the operation of the damping cylinder in a prosthetic leg proceeding through its walking cycle (6). This jig consisted of a variable speed d.c. motor which turned a

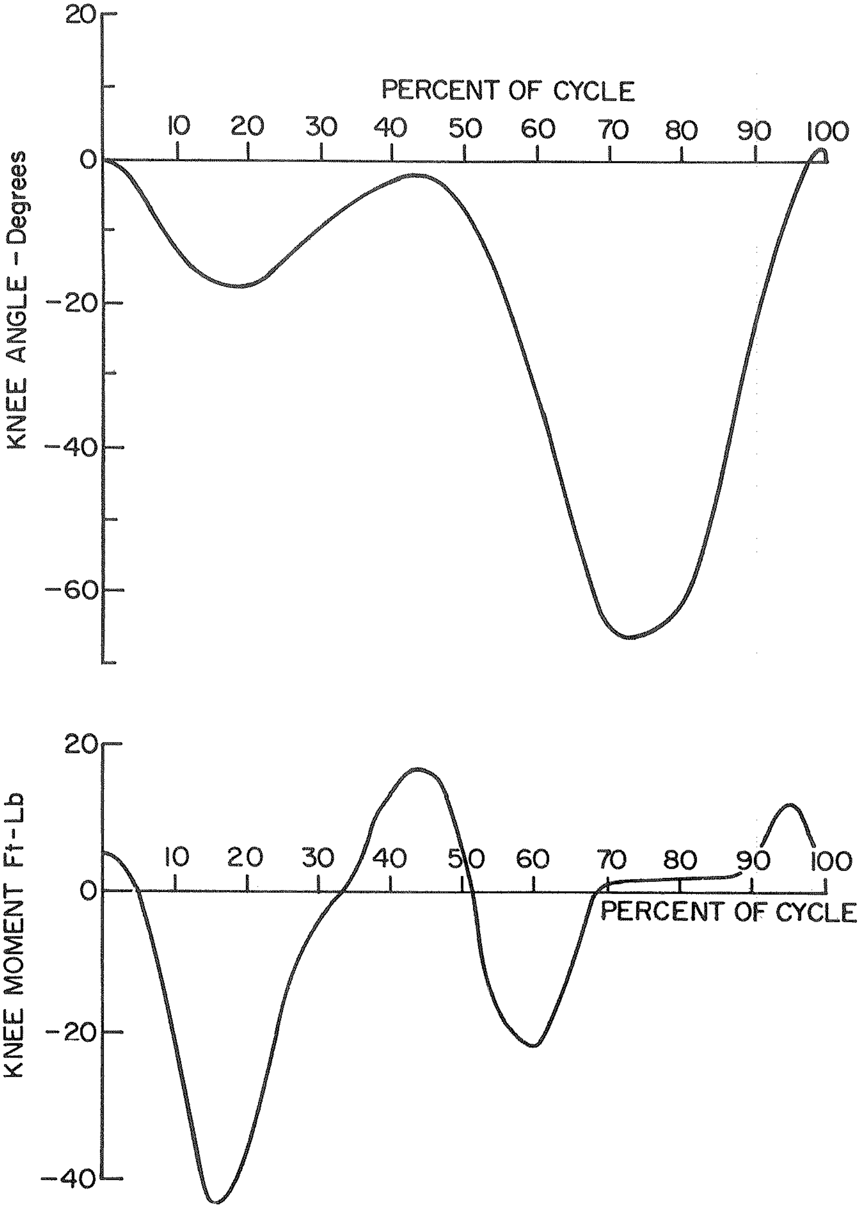

FIGURE 3.—Typical knee flexion and bending moment about the knee during normal human locomotion.

driving cam to simulate walking at different speeds. The cam, which was designed to produce the appropriate piston velocity versus percent

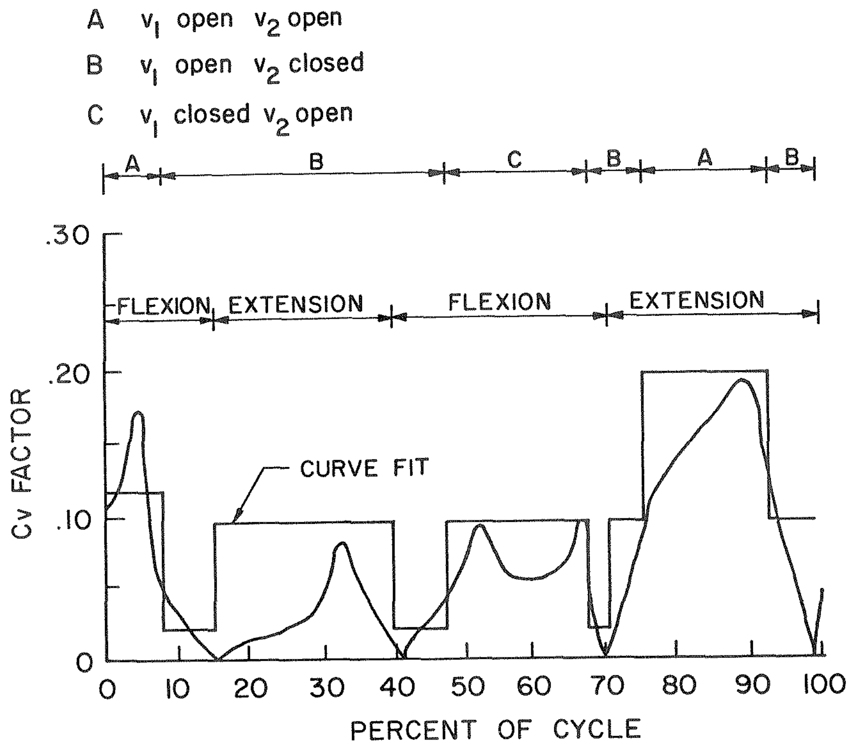

FIGURE 4.—Desired Cv factor versus percent walking cycle and the corresponding Cv curve fit.

walking cycle in the damping cylinder, activated this damping cylinder via a lever. A second cam was mounted on the same shaft as the driving cam to activate microswitches which energized the bypass valves in a predetermined sequence. Strain gages were mounted on the piston rod of the damping cylinder to measure the force versus percent walking cycle in this piston rod.

The first tests on the jig consisted of operating the system at speeds corresponding to walking rates of 40, 60, 80, and 90 steps per minute (SPM) and recording the forces generated at each speed. A time reference on these recordings was established by recording the switching voltage of bypass valve V2 simultaneously with the forces. A typical recording is similar to that shown in Figure 5. Inspecting the forces recorded at the various SPM, it was observed that their relative behavior was as expected; i.e., higher force magnitudes were observed at higher values of SPM.

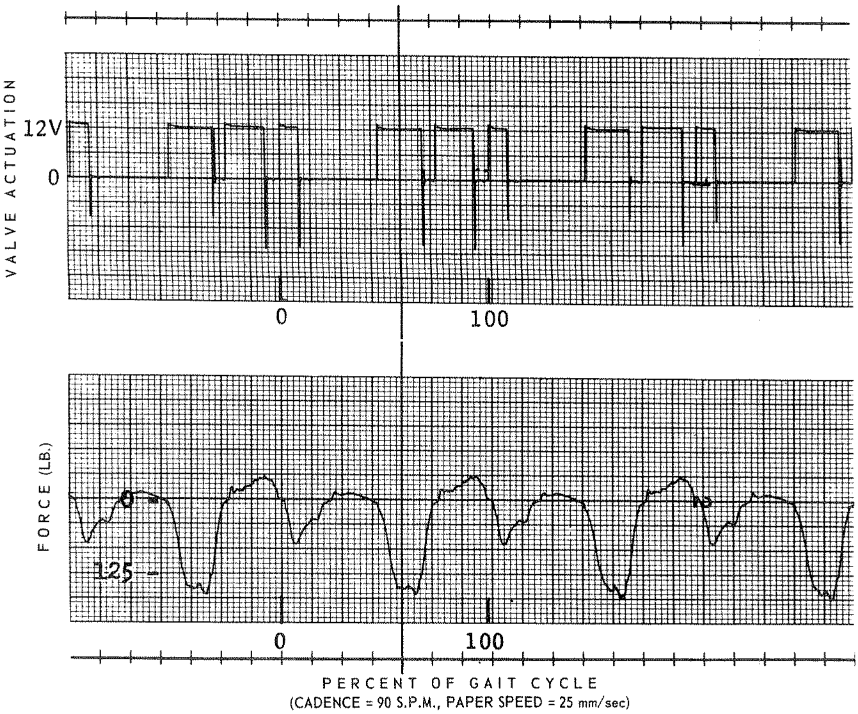

FIGURE 5.—Force (in pounds) versus percent walking cycle as measured via strain gages on damping cylinder piston rod.

Since locomotion data exist only for normal walking speeds of about 90 SPM, comparisons between the measured and previously established forces in the damping cylinder can be compared at this one speed only. It was found that the measured force following heel contact was low. To rectify this, the switching of bypass valve V2 at heel contact was eliminated to increase the resistance at this point (Fig. 6). The resulting forces are shown in Figure 7, where the measured force is off the test jig and the physiological force is that established from locomotion data. It is seen that these two forces compare well, considering the fact that the physiological force curve is derived from normal locomotion data which resulted from muscular activity. The measured force curve of the damping cylinder is derived from a purely passive element; i.e., the damping cylinder cannot supply power to the knee, it can only dissipate it. Because of this, the hydraulic system needs a nonzero flow in a given direction to produce a force in the opposite direction. For example, just after heel contact the flow is in a direction (Fig. 3) producing a resistive force in the negative direction. Then a continuing force in the negative direction must be supplied by a flow in the reverse direction. This is impossible with a passive element. Therefore, this prosthesis will need some active input from the stump to maintain a normal gait.

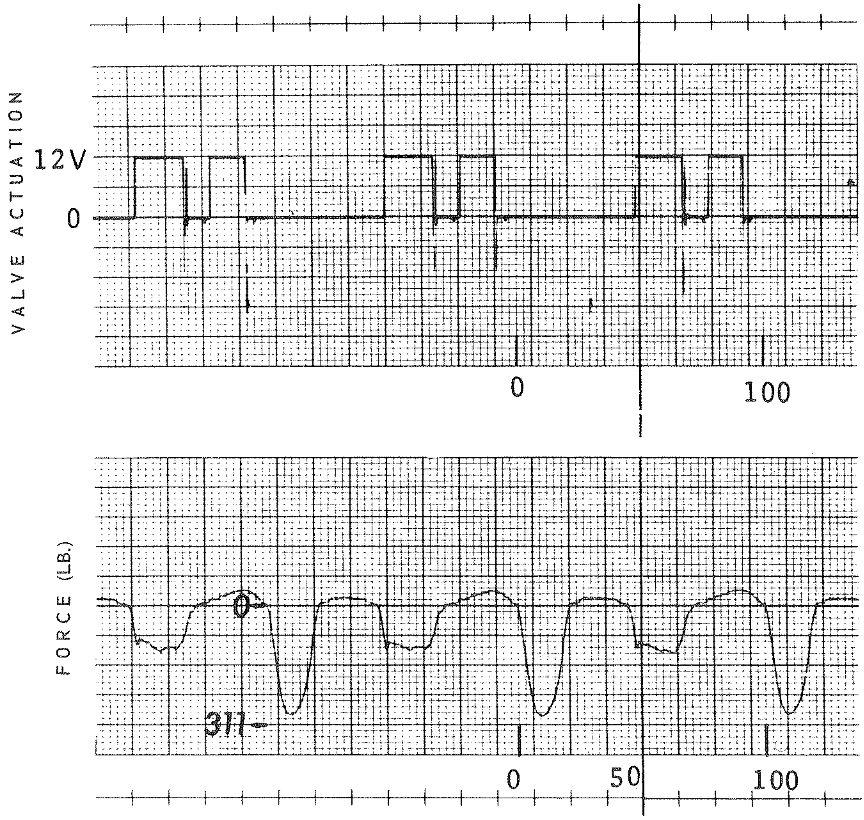

FIGURE 6.—Measured force (in pounds) on damping cylinder versus percent walking cycle.

To this stage of development, the hydraulic system required a four-state control system, i.e., four possible levels of resistance available during one walking cycle. Therefore, three discrete levels of muscle contractions are implied, a fourth state corresponds to rest. In anticipation that four states of differentiation may be too difficult for the amputee to achieve, a three-state control system was investigated. It was implemented by eliminating the mode where valve V1 is closed and V2 is open (mode C in Fig. 4), and by replacing it with the mode where both valves are open, resulting in a slight change in the Cv schedule. Retesting

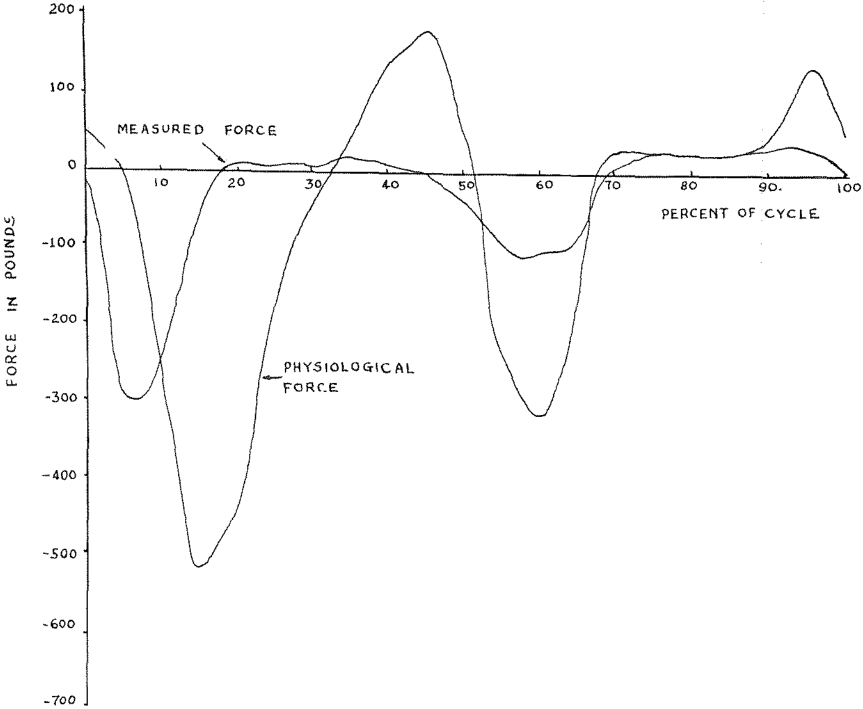

FIGURE 7.—Measured and physiological forces.
Measured force—empirical values obtained from tests.
Physiological force—analytic values calculated from data obtained in normal locomotion studies.

the hydraulic system with this modification showed that the change in the force versus percent walking cycle was also small (maximum difference was less than 15 percent). Therefore, the three-state control system was considered an acceptable alternative to the four-state control system, if necessary.

Other tests performed on the test jig were made to determine whether there existed any significant changes in force wave shapes and magnitudes if valve switchings were altered slightly, as might occur from inadvertent signals or activation by the amputee. No noticeable problems appeared. The hydraulic system was then fitted into an existing prosthesis in preparation for amputee testing.

DESIGN OF EMG CONTROL ELECTRONICS

The EMG electronic controller consists of a circuit which uses EMG signals as an input and, depending on the level or strength of contraction, establishes the desired resistance to flow. Since EMG signals consist of alternating signals of a few millivolts, a high gain amplifier and envelope detector were needed to feed a set of three comparators with different threshold voltages. The outputs of the comparators activate a logic circuit, which in turn energize the correct valve combination. Included in the design is the provision for conversion to a three-state control system, in case the amputee experiences difficulty in controlling four states.

Before proceeding with the discussion of the design, the logic of operation of the four states must be considered. It is known that during stance phase there is quadriceps and hamstring activity to attain stability during weight-bearing, whereas during swing phase muscular activity is at a minimum. Knee locking would normally occur in a stumbling situation, aiding recovery, but resulting in a high degree of muscular activity. These facts led to the following as a logical sequence for the different control states. The first, or rest state, would consist of both valves open for a free and easy swing and the fourth state would consist of both valves closed for a lock to flexion. States two and three would consist of only one or the other valves open with increasing resistances, respectively.

A schematic of the circuit is shown in Figure 8. In Block 1 of this circuit are the input or buffer stages which have unity gain, and provide a high input impedance for the differential muscle signal (59.4 MΩ common mode, 10 MΩ differential mode). Since paste electrodes were used, these stages provide good isolation. The outputs were connected to an adder which included an input impedance balancing potentiometer. When adjusted, this potentiometer established a −94 dB common mode rejection (CMR) at 60 cycles. To stabilize this CMR, the buffer stages were a.c. coupled to the adder with two matched capacitors to isolate d.c. drifts at the outputs of the buffers.

Block 2 consists of a two-stage variable gain (5,000 to 15,000) amplifier and envelope detector. In Block 3 the output of the envelope detector is connected to three comparators. These comparators, with three different reference voltage levels, can distinguish three different output levels of the envelope detector. With the resting level as an additional state, this results in a four-state controller. In addition, a hysteresis effect is provided for at the comparators to avoid valve chatter.

In Block 4 the outputs of the comparators are sensed by a simple logic circuit which, depending on the comparator states, energizes one or both power transistors connected to the logic circuit. These transistors are used as power gates to the two solenoid valves in the hydraulic system. In addition, Block 4 includes a provision for easy conversion to a three-state control system. This is accomplished by breaking the connection between comparator A7 and the diode by a single-pole-double-throw switch which connects the diode to either the comparator output (four-states) or ground (three-states). When the diode is grounded the logic circuit can only be activated by comparators A8 and A9.

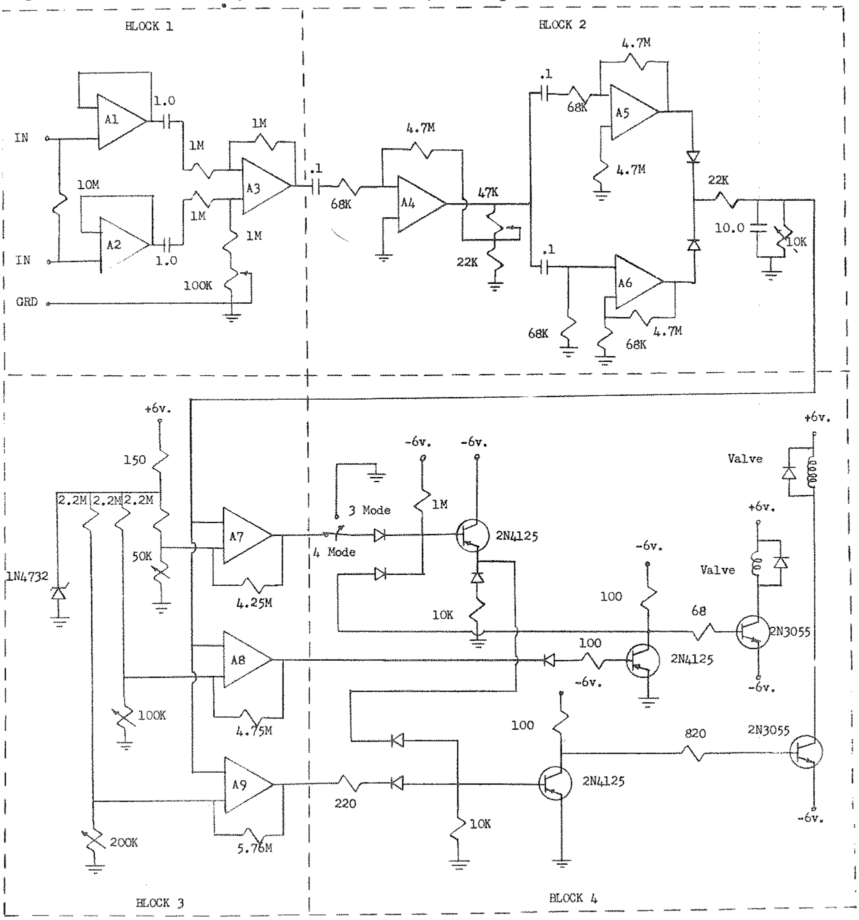

FIGURE 8.—Circuit diagram of EMG control electronics.

Ten nickel cadmium rechargeable batteries (1.25 v. each) formed the portable power supply for the solenoid valves and the electronics. The batteries were rated at 1.2 A.-hr.

The circuit was bench tested using the EMG signals of a forearm muscle. The results proved that the circuit could be successfully operated (three or four states) with just a few minutes of training. Figures 9, 10, and 11 show a typical EMG signal before any processing, after full-wave rectification, and after envelope detection.

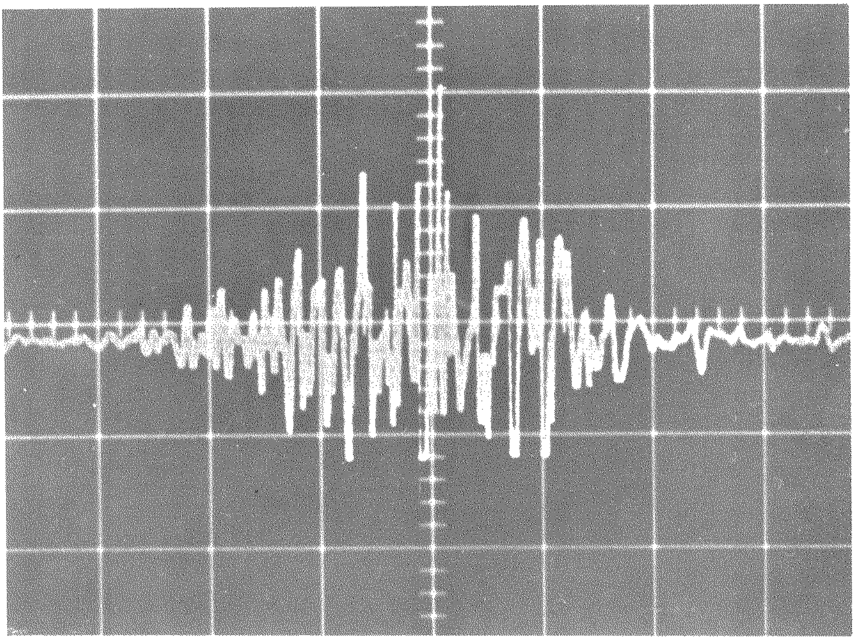

FIGURE 9.—Typical EMG signals before processing. Vertical scale is 2 v./cm., horizontal scale is 0.1 sec./cm.

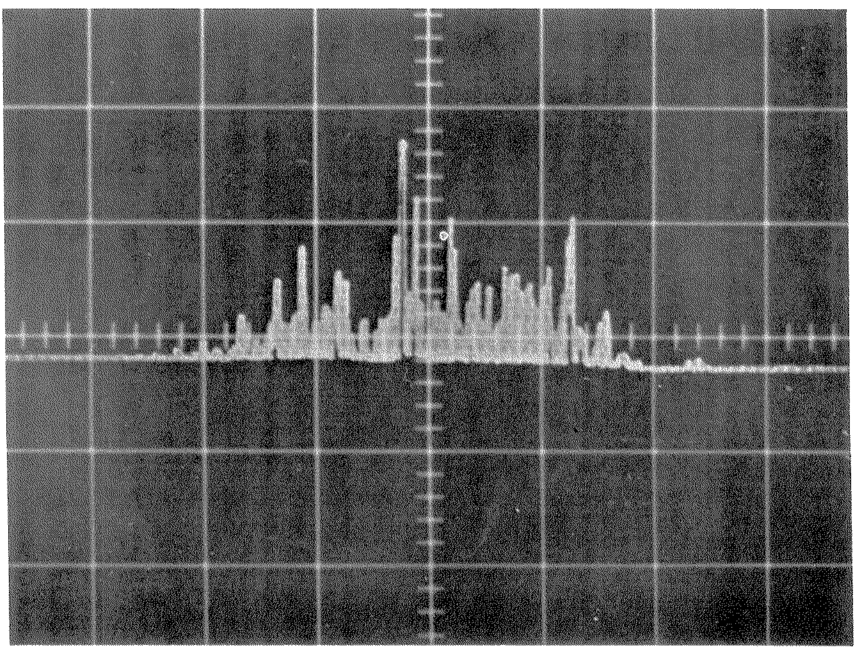

FIGURE 10.—EMG signals after full-wave rectification. Vertical scale is 2 v./cm., horizontal scale is 0.1 sec./cm.

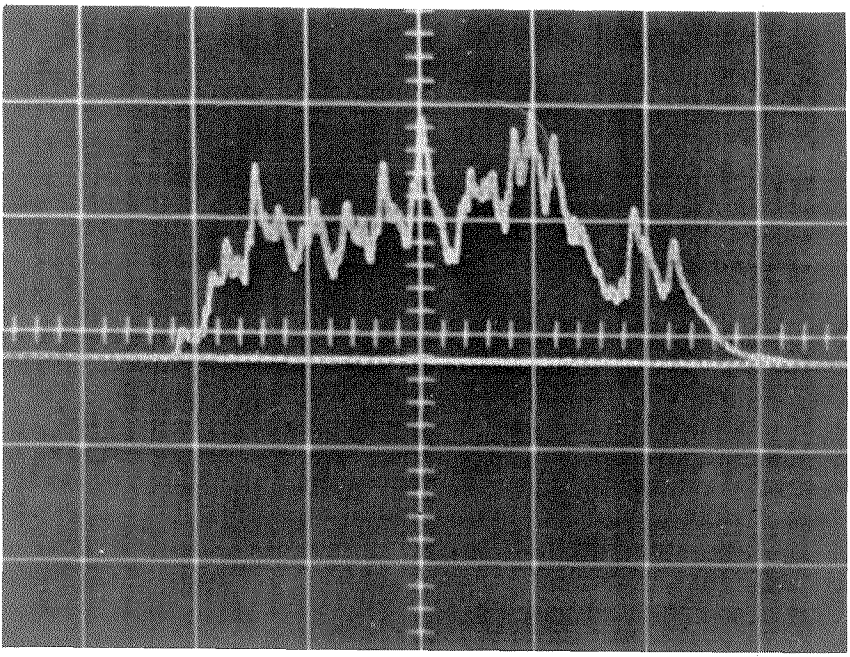

FIGURE 11.—EMG signals after envelope detection. Vertical scale is 0.1 v./cm., horizontal scale is 0.1 sec./cm.

AMPUTEE TESTING

The four-state control system was mounted on a belt along with the power pack, and the hydraulic system was incorporated into a prosthesis (Fig. 12 and 13). To determine what states were being operated by the amputee, two low-power indicator lights were connected across the solenoids and mounted on the belt for easy viewing. The amputee, who was amputated in 1965 about 6 in. above the knee, was then asked to test the electrohydraulic system (Fig. 14). The results were as follows: Although the prototype weighed 10 lb. and the amputee's standard hydraulic prosthesis weighed 7½ lb., he did not find this a deterent. However, when walking, the knee joint was too viscous using hydraulic oil to operate properly, so kerosene was used instead.

Initially, paste electrodes were placed on the rectus femoris, because it possessed the EMG signal of greatest strength. When seated, the amputee was able to control the four states quite well after a training period of about 15 min. However, during gait the threshold level on comparator A7 had to be raised because of the high resting noise level. The necessary lowering of the threshold level on comparator A9 to permit

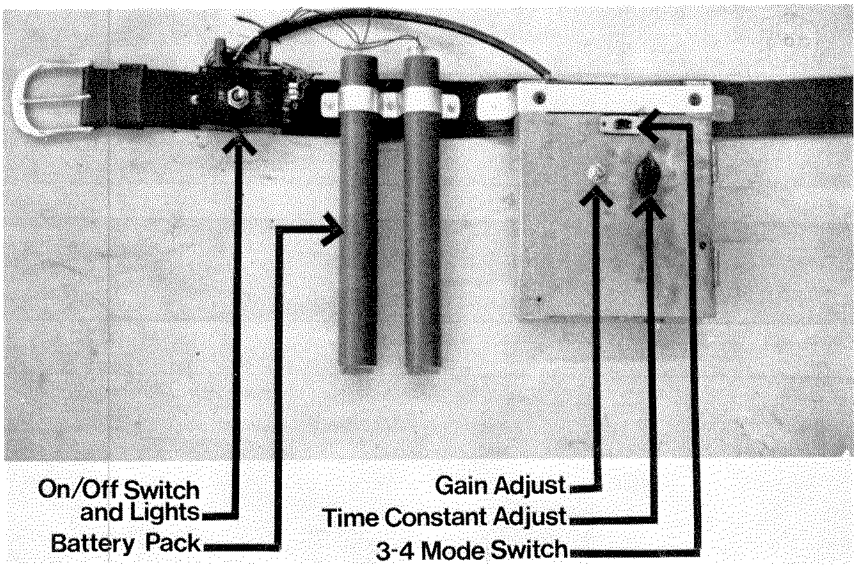

FIGURE 12.—Prototype of electronic control unit.

locking also reduced the active range, and the amputee had difficulties controlling four states accurately because of their close proximities. Therefore, it was decided to use three-state control in future tests with the amputee.

A problem was also encountered with the site of EMG pickup. Although good control was achieved by contractions of the rectus fermoris while in a sitting position, during ambulation when the hip was flexed an inadvertent locking of the knee mechanism consistently appeared at given times during the walking cycle. This implied that the rectus femoris was still being used far too actively in hip flexion and therefore could not be used as an EMG site for indepent, voluntary myoelectric control. Since the amputee was using his hamstrings to a far greater degree than his quadriceps for prosthetic knee stability, the electrodes were mounted into the socket wall over the semimembranosis.

After several sessions, to solve minor problems in the electrohydraulic system, the system was successfully tested. Within a few minutes of training the amputee was able to control the three states quite accurately. Minor adjustments to comparator threshold levels were required, followed by a short training period. The amputee quickly adjusted to the new settings and was able to control all levels of knee resistance under normal walking conditions.

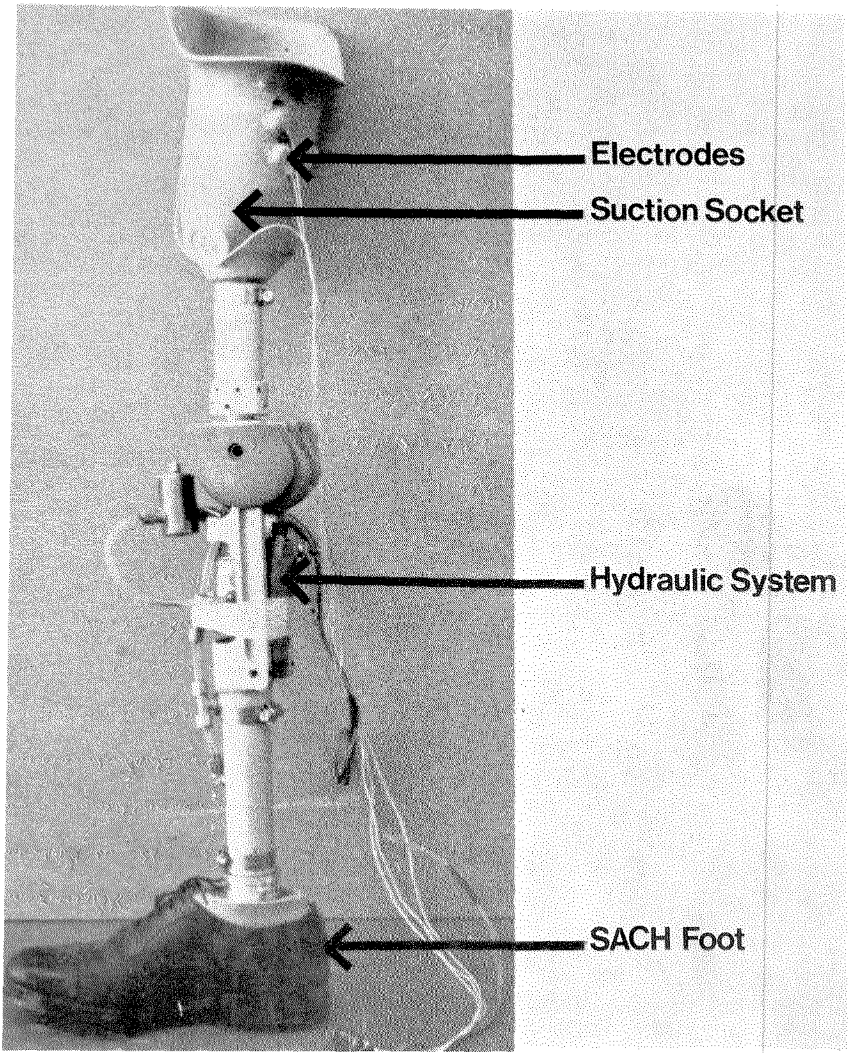

FIGURE 13.—Voluntarily controlled electrohydraulic prosthesis prototype.

The amputee was then asked to operate a specific state at a specific point in the walking cycle. This was not natural for the amputee and he tired quickly, but the results showed that this was possible. Another test performed was having the amputee walk normally, i.e., not consciously trying to activate any particular state. This was done to discover how naturally the leg was being operated by the amputee, by observing via the lights on his belt what states were being activated at the different phases of the walking cycle. The results showed that the correct states

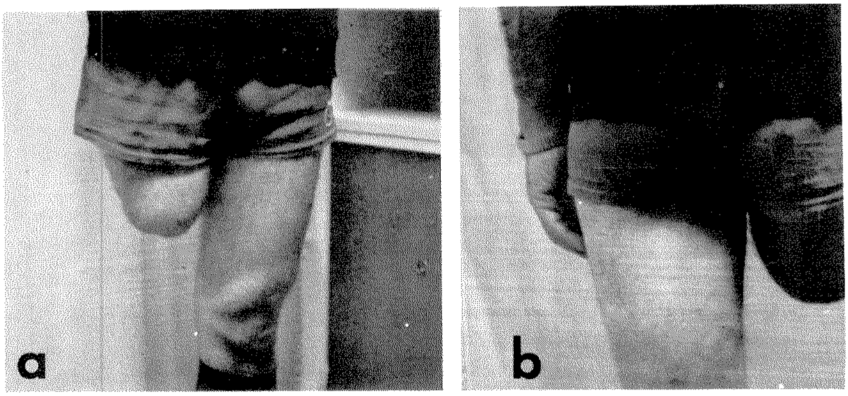

FIGURE 14.—a. Anterior view of amputee, and b. posterior view of amputee.

were being operated automatically at the right times, which led to the anticipation that this system required minimal extra training.

At one point during the trial an inadvertent stumbling situation was encountered. The amputee automatically recovered by locking the knee and preventing a fall.

Gait studies were conducted using the locomotion study facility at the Shriners Hospital, Winnipeg (7). From the resulting kinematic and kinetic analysis the knee joint torque and the net powerflow from or energy dissipated by the prosthesis during the swing phase were found. From the analysis of one stride the average joint torque was approximately 2.5 N-m (dissipative) resulting in a power flow of about −5 w.

Considerable work remains to be done to make this a practical prosthetic system. This includes: reducing the size and weight of the electrohydraulic system, adding a biofeedback mechanism which would give the amputee a sense of prosthetic knee position, and providing dry electrodes instead of wet electrodes, thereby minimizing preparation and maintenance.

CONCLUSION

A voluntarily controlled variable resistance lower-limb prosthesis was developed from a detailed analysis of normal gait. Control is voluntary since the input to the control electronics comes from EMG signals derived from residual thigh muscles in the stump. A prototype was built consisting of a standard endoskeletal prosthesis fitted with a hydraulic damper across the knee, bypassed by two solenoid valves. The system was tested on an amputee who voluntarily controlled via EMG signals the damping resistance across the knee with a discrete three-state control system.

The amputee was very pleased with the performance of the prosthesis ecause it was natural to operate. He was especially pleased when he utomatically locked the knee in an inadvertent stumbling situation ncountered in the laboratory and was able to recover from a potential all.

The successful results of this project demonstrate that the concept of n EMG voluntarily controlled hydraulic prosthesis is viable. However, t is recognized that further research will be required to make this a ractical system for above-knee amputees.

ACKNOWLEDGMENTS

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