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A KINEMATIC EVALUATION OF AN EMG-CONTROLLED ABOVE-KNEE AMPUTEE PROSTHESIS

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ABSTRACT

Rehabilitation of an above-knee (A/K) amputee relies upon a well-designed prosthesis to give him functional support, and aesthetic posture during ambulation. To evaluate such a device requires thorough understanding of the biomechanics involved and extraction of kinematic information.

The success of electromyographic (EMG) control in powered upper-limb prosthesis has prompted the use of EMG control in a lower-limb prosthesis which varies the amount of viscous damping across the knee joint. In a former project, an EMG-controlled prototype was built, and positive results were noted.

In this study, the hydraulic and electronic designs of the original prototype were improved and modified to make it suitable for long term testing. This was followed by extensive evaluation which includes data acquisition and analysis on several experimental runs conducted during a training period. The performance of gait is compared to available data.

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CHAPTER 1

INTRODUCTION

Physiological deficiencies as a result of trauma, disease, congenital anomalies, and aging can lead to loss of lower or upper extremities. Replacement of the extremities requires prosthetic devices which can eventually duplicate the normal function of a healthy knee joint.

The case of an unilateral above-knee (A/K) amputee deserves extra attention because it involves factors common to all lowerlimb amputees. It provides perhaps the best example in the locomotion study of prostheses.

The majority of A/K prostheses in use today employ mechanical friction in the knee joint design, which does not permit the most efficient or normal appearing gait. Improvements have been achieved with mechanically 'programmed' hydraulic and pneumatic units; that is, the amount of damping resistance in the units is predetermined. However, functions like varying cadence, and walking up or down stairs, on ramps, and over rough terrain still cannot be performed in a normal manner. The use of myoelectric signals to control knee actions has been demonstrated at least to some extent by Horn (1972), who designed a knee unit which can be locked to flexion voluntarily by the wearer and is unlocked automatically at full extension, allowing for free swing. This design allows for weight bearing on a flexed knee but does not allow for varied cadence.

An EMG-controlled A/K prosthesis has been developed at the Univ. of Manitoba based upon data from normals. A hydraulic damper was built inside the knee unit, which can vary the knee action from 'free swing' to 'lock' and from 'lock' to 'free swing'. This variation is achieved by controlling the resistance of fluid flow around a damping cylinder with bypass valves through electromyographic (EMG) signals from the hamstrings of the stump (Dyck, 1974). A schematic of the prototype is shown in

Fig. 1.1. It consists of a hydraulic damper, whose resistance is varied by the control of two solenoid valves and one check valve. The check valve opens only during the extension phase, as locking is desirable only in flexion.

The mechanical and electronic portions of the prototype built in Dyck's work was bulky, and the electronics did not work reliably. Therefore, a new improved version was designed. This new version was then benched tested and tried by the amputee.

The main purpose of this study is to perform extensive evaluation on the new unit. To do this, the amputee experimented with it once every other week for approximately six months. During this period, locomotion runs were monitored via an existing video-computer system (Winter, et al, 1972). Based upon the kinematic information extracted, velocity and acceleration profiles of the limb segments could be obtained. When these profiles are coupled with anthropometric data, parameters such as force, torque, and energy levels could be calculated. Necessary computer programs were rewritten to accommodate the analysis of asymmetrical gait. The performance of the amputee-prosthesis system is then compared and analyzed for different control conditions, and tasks such as walking on ramps and climbing stairs.

It was very difficult to establish concise criteria for evaluation because of the large amount of data and the number of variables present. In spite of these difficulties, results of this study strongly indicate that a controlled EMG pattern can be conditioned through training in such a way that subconscious control is possible. Performance of the amputee on this prosthesis was found to be satisfactory.

CHAPTER 2

BACKGROUND

(I) Walking cycle

Essentially, a walking cycle consists of a stance phase, which constitutes approximately 60% of the cycle; and a swing phase, which constitutes approximately 40% of the cycle. The stance phase begins when the heel makes contact with the ground (HC) which is positioned as the 0% mark. The foot flattens (FF) at about the 20% mark; then by the plantar flexion of the foot and the initiation of knee flexion by the forceful action of the calf muscle, the body is propelled forward as the heel leaves the ground (HO) at about the 40% mark. The stance phase ends when the entire foot rises from the ground. The swing phase begins with toe-off (TO) at about the 60% mark, and ends when the heel strikes ground at the 100% mark again.

At the end of the swing, there is a period of deceleration when the forward motion of the leg is restrained by the stretching of the hamstring muscles to control the position and velocity of the foot immediately before HC. The period when both legs are on the ground is called the double support phase (DS). These parameters are shown on Fig. 2.1. The duration of the DS period is related to cadence, and the shortening or the absence of DS indicates that a person is running rather than walking.

(II) The general walking pattern of A/K amputees

It appears that in walking, the body is constantly converting energy from potential to kinetic and then back to potential energy again (Inman, 1965). The path of centre of gravity (CG) has been described as the intersection of the horizontal and vertical displacements shown in Fig. 2.2 during level walking of normals (Saunders, 1953). That is to say, the CG is continuously displaced sinusoidally in both vertical and horizontal directions. Therefore normal gait is harmonious, and energy among body parts can be balanced and transferred.

The gait pattern of A/K amputees varies considerably from normal. An A/K amputee has to elevate his body in order to keep the harmonious path of CG just before HC of the artifical leg when the trunk is descending. If the landing leg were the normal leg, such action could be accomplished by planter flexion of the ankle and flexion of the knee. Due to the deficiencies present in both ankle and the knee of the artificial leg, a different pattern results in angulation of the natural knee and ankle, or the tilting of pelvis (see Fig. 2.3; Eberhart, 1968).

Until the prosthetic heel rises, the leg rotates forward about the ankle joint of the foot. The knee is consequently in an extended position at the time the hip joint reaches its maximum elevation. This increases the vaulting sensation.

At this time, the hip joint is descending while the prosthetic leg is still on the ground. As it is unable to control stability, the normal leg is called upon for aid by landing earlier on the ground.

During the swing phase, the forward swing of the normal leg is initiated by powerful contraction of the plantar flexor of the ankle. Lack of such a mechanism in the prosthetic leg forces the amputee to rely heavily on his hip muscules for initiating the forward prosthetic swing. Fortunately the hip muscles are under voluntary control and consequently can adapt themselves within limits.

The lack of muscular control at the knee joint severely reduces its ability to create the normal knee moments. The A/K amputees compensate by shortening the steps in order to reduce the moment developed to flex the knee (Peizer & Wright, 1974). The inability of the knee to flex during early stance phase increases the amount of energy required to walk. This mainly results from the excessive rise of centre of gravity during the prosthesis stance.

(III) Basic requirements of A/K prosthesis

Essentially, an A/K design requires that the prosthesis does not collapse during the stance phase. This does not mean that locking the knee throughout the entire stance phase is optimum. In normal gait, the knee is fully extended at heel contact. It then flexes slightly (accompanied by planter flexion) and re-extends during weight bearing. This action allows the hip to follow a more harmonious path and therefore requires less energy. Conventionally, prostheses are aligned in such a way that the centre of rotation of the knee is dorsal to the load line (see Fig. 2.4), so as to prevent the knee from collapsing during weight bearing. This alignment applies a hyperextension to the hinge, locking it against the stops.

During the swing phase, an A/K prosthesis can be considered as a pendulum subjected to driving forces by the hip and thigh. At the beginning swing, a forward acceleration acts on the upper part of the prosthetic leg (socket portion and thigh muscles). If no resisting moment is supplied by the knee mechanisum, an excessive heel rise will occur. Also, at the end of swing, the shank will have a rotational velocity and must stop abruptly. Both problems become more serious as walking speed increases (see Fig. 2.5). Therefore, the minimum requirement is that some damping must be provided during the swing phase.

One thing to note is that, during the double support phase, the energy of forward translation of human body comes from a force couple, that is, from the push-off and the deceleration of the swinging leg. The ratio of the components from push-off and deceleration of the swinging leg has been found to be approximately five to eight (Inman, 1965). This phenomenon explains that if a prosthesis is too light, the amputee will have difficulty in developing sufficient kinetic energy at the end of the swing phase to be fed back into the system to maintain the forward velocity of the body.

Other factors that require consideration are: safety and reliability; flexibility for sitting; whether it is cosmetically acceptable; and if the energy cost is at a minimum.

(IV) Designs for A/K prosthesis

A) Constant friction

This provides friction in the knee bolt, which rotates with the shank, and a mating surface fixed to the upper leg. Variation of the resistive torque is adjusted by the contact force in the brake. "Bumper" springs have been provided to help control deceleration at the end of swing. This system provides a relatively constant resistive force independent of walking speed and angle of flexion.

B) Intermittent friction

This provides a resistive torque that is a function of the knee flexion angle. Mechanical friction is provided by the pressure between the three disks mounted concentrically with the long axis of the knee bolt. The resistance offered by each individual disk is varied by a wheel during different intervals of the swing phase. Essentially it provides a better torque profile than the constant friction type. It functions best only at one cadence and is purely dissipative.

C) Fluid control designs

Recently, mechanisms having orifice-flow damping have received much attention. The Henschke-Mauch (Mauch, 1968) and Dupaco "Hermes" units are well-developed examples. They contain flow channels and orifices in the piston cylinder, so that the moving piston can sucessively block off escape of the fluid and thus vary the resistance throughout the swing phase in order to approximate the 'ideal' moment curve. Check valves switch fluid flow to separate channels for extension and flexion, thereby allowing independent adjustment of flexion and extension resistance.

D) Pneumatic control design

An air dashpot can be used as a resistive element. The main advantage is that fluid leakage is not a problem and therefore the cost of manufacturing is lowered. The air in the cylinder serves both as an energy dissipator and a spring.

E) Polycentric knee

The most promising mechanism at present uses four-bar linkages together with fluid damping (Radcliffe, 1957). Since the action of a normal knee joint is also polycentric, the four-bar linkage provides different instantaneous centres (usually are very small distances apart) with relative changes of the positions of the limb segments.

Besides satisfying the stability requirement as described by Radcliffe (Radcliffe, 1957), the polycentric action also contributes somewhat to the sense of knee position as in normal individuals.

All of the mechanisms described above provide predetermined amount of damping resistance, and the control is passive.

F) Myoelectrically controlled knee locking mechanism

The design by Horn (Horn, 1972) uses myoelectric signals to lock the knee at any desired angle. Once locked, the knee can only extend, and unlocks itself at full extension. The advantage of this system is that a wearer can bear weight on a slightly bent knee, allowing for a more normal appearing gait. Horn also claims that it can be used for climbing stairs. This design does not, however, allow the amputee to vary the resistance as a function of the knee angle, nor to control this resistance in order to accommodate different gaits.

CHAPTER 3

EXPERIMENTAL DESIGN

(I) Hydraulic system

The hydraulic system of this study is a modification of the work done by Dyck (Dyck, 1974). A hydraulic damper with a discrete damper control is considered adequate to be implemented with a proportional myoelectric controlled scheme.

In Dyck's work, data from normals was gathered to determine the required characteristics for this hydraulic system, such as the pressure rating of each component, the rates of flow, the resistance to flow versus percent walking cycle, and the number of bypass valves and their resistance. Also in Dyck's work, with the knowledge that a valve's resistance to flow is characterized by its C_v factor (which is defined as the flow of 60^{o}F water in gal./min. through the valve with one p.s.i. pressure drop across it), the desired C_v needed across the damping cylinder was calculated and adjusted for the desired hydraulic fluid.

By curve fitting, he then determined the number and type of valves needed . Fig. 3.1 shows his resultant C_v curve versus walking cycle plot which calls for two solenoid bypass valves and one check valve, all in parallel with the damping cylinder. Dyck's design used two solenoid valves, V1 and V2, with C_v factors of 0.02 and 0.1, respectively, while the check valve has a C_v factor of 0.08.

The hydraulic prototype used in Dyck's project was made up of several individual components - a hydraulic cylinder, two electrical actuated valves, and a check valve. The hydraulic lines were connected together externally using elbows, unions, and plastic tubings as they were intended for temporary usage. In this study, it was felt that a more compact and permanant assembly was required for a prolonged period of training. Therefore, a new version of the hydraulic prototype was designed to reduce the bulkiness of the unit. It has all the components mounted into a single unit: the check valve was built into the piston , while two new bypass solenoid valves were mounted on the side (in parallel with the hydraulic cylinder) to be controlled by EMG signals.

The schematic and appearance of the new hydraulic unit are shown in Figs. 3.2 and 3.3 respectively. These figures also show two external valve screws in which adjustment can be made on the size of the orifices of the two bypass valves, either to increase or decrease the resistance to flow.

With this arrangement, the amputee is given a wider choice of resistive damping settings by selecting different combinations of the closure of the two bypass valves and the check valve according to his needs. With the maximum settings on the two bypass valves, the Cv factors for the designated High Resistive Valve (HRV), and Low Resistive Valve (LRV) are approximated to be 0.04 and 0.14 respectively, with the Check Valve (CHV) at 0.16 (see Appendix A). A coil spring was added to the top of the cylinder to provide energy storage which supplies an extra lift during push-off.

In the rest state, both solenoid valves are normally open; and while standing, the amputee will achieve his stability by prosthesis alignment. In the flexion mode, since the CHV is inoperative and since the two solenoids have different C_V factors, there exist three additional controllable levels of resistance - one of which is 'lock'. During extension, all corresponding levels of resistance will be decreased due to the release of the bypass CHV. These combinations (Fig. 3.4) constitute the same 'four-mode' operation as described by Dyck (1974).

(II) Myoelectronic system

The basic concept of myoelectronic control is that when a muscle contracts, the contraction is accompanied by an electrical signal, called a myo-electric or electromyographic signal (EMG). This signal can be detected by placing electrodes on the surface of the skin. One of the most useful characteristics of the EMG signal is that its root-mean-square (RMS) value is proportional to the isometric contractal force of the muscle involved. An EMG signal processor had been designed by Dyck (Dyck, 1974) to control the operation of solenoid valves utilizing such a characteristic. However, repairs on this circuit were needed constantly because of a poor wiring technique was employed. Therefore it was decided that a new printed circuit (PC) was to be built with a reduced size in mind. A block diagram of this processor is shown in Fig. 3.5, which is basically derived from Dyck's design. Also, the new circuit is shown in Fig. 3.6.

The typical EMG signal has peak amplitudes from 100 mV to 4mV and lies in the frequency band of 20-500 Hz. Therefore, it requires high gain amplification and 60 Hz.

noise rejection at the front end. The first stage of the processor consists of two high input impedance instrumentation amplifiers (AD 521, a data sheet is provided in the Appendix B). The first one acts as a low gain differential amplifier (with a gain of 20), which has a built-in feature of high common mode rejection ratio (CMRR) rated at 90 db to 60 Hz noise. As the skin-electrode interface generates DC potentials which may be up to 100mV in magnitude, it is filtered out by means of a high-pass filter with a corner frequency (f_c) of around 15 Hz, in order to prevent saturation before entering into the second high gain amplifier. The second amplifier has an adjustable gain of 100 to 1,000; therefore the total amplification can be set from 2,000 to 20,000.

The second stage is a full wave rectifier, the output of which is a signal whose DC component is proportional to the EMG signal's RMS value. To obtain this DC component, it is fed through a second order low-pass filter (LPF). The output from the LPF is a DC voltage that corresponds to the level of muscular tension.