Intelligently Controlled Above Knee (A/K) Prosthesis

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ABSTRACT

In existing advanced A/K prostheses the stiffness and damping at the prosthetic knee joint are applied by hydraulic or pneumatic cylinders. The pre-set parameters being constant, are not optimal during the whole gait cycle, for different walking speeds, and sudden posture irregularities. The complexity of normal and prosthetic gait leads to the development of 'soft' control (non-analytic) governed by a finite set of rules. The pre-defined and stored parameters and rules are combined as the knowledge-base of the system. This knowledge is based on existing experience and known facts about both normal and prosthetic gait. The methodology of developing the controlled prosthesis was first to study normal gait by modeling and simulating level human gait. Then the prosthetic gait was investigated using the normal gait model with a passive (parameter controlled) and a realizable knee joint. The results of the prosthetic gait investigation gave the values and ranges of the controlled impedance parameters, the predicted performance of the controlled prosthesis, and the moment (effort) required from the amputee. A laboratory prototype of a controlled A/K prosthesis was built and evaluated in the clinic.

INTRODUCTION

Lower limb prosthetics is aimed in restoring the mobility of an amputee. This is an essential element in the rehabilitation process done by replacing the missing limb (or a part of it) by an artificial one. Prosthetics has been able to improve the mobility of lower limb amputees by offering a wide variety of prosthetic components and mechanisms, but not yet to a satisfactory degree, and the gait of an A/K amputee still has noticeable abnormalities. The motions of the body segments serve for locomotion and maintaining stability. During one gait cycle, the leg configuration performs two different phases, twice. One is the Single Limb Support phase (SLS) when one leg is swinging (almost 'freely') forward and the other leg supports the body. The second is the Double Limb Support phase (DLS) when the two legs stand on the ground and both support the body.

The knee has a significant function in human (legged) locomotion, providing features such as bearing body weight, absorbing the impact of the body with the ground, maintaining posture stability, contributing in initiating swing, and in body lift off in slope and stair climbing. When the leg is hanged at the hip and swings forward, the knee flexes to keep a small ground clearance and then rigidly extends to accept the body weight. When the leg supports the body, reaction forces at the hip and at the foot act on the leg. The knee, supported by the muscles, supply these reactions. The knee then also flexes to some degree to prevent high vertical rise of the body.

A controlled A/K prosthesis is an artificial limb attached to the amputee's body below his hip and its control system has to cooperate with the amputee's biological locomotion system. An ideal A/K prosthesis is expected to perform with the requirements of preserving sufficient stability to support the body weight, a capability to absorb the ground impact at heel contact, to allow smooth forward progression with sufficient knee flexion, and to have instant response to a change in walking speed, mainly during swing. The potential of a controlled A/K Prosthesis lies in the properties of automatic safety reaction by locking or damping the knee flexion during loading, a wider range of activity for the amputee with a single axis knee joint, control over the whole gait cycle of stance and swing, sitting, standing, and stairs/slop walking modes, optimum knee damping and stiffness, adaptation to walking speed and amputees parameters with on-line speed identification and application, and inherent diagnostics and performance evaluation with a computer supervision.

The main task of the present work was to develop a self contained controlled prosthesis with sensors attached to the prosthesis only, to feed back a time varying 'soft' control command through the whole gait

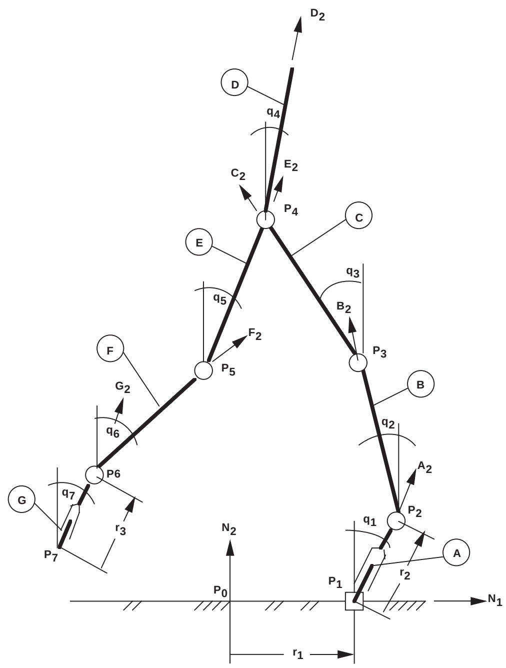

Figure 1: The seven link model in SLS viewed in the sagital plane

cycle. The objective is to develop a new systematic method for implementing a 'soft' control which is required to be expandable, that will reach the functional and operational requirements, and will guarantee robustness.

The following presents the complete development process of a controlled A/K prosthesis. The first section describes the analysis of normal human gait based on a bipedal gait model. The second section presents the analysis of a (above knee) prosthetic gait during swing and stance phases of gait. The third section presents the synthesis, implementation, and evaluation of a laboratory prototype, of a controlled A/K prosthesis. The conclusions and summary are presented at the end of the paper.

MODELLING NORMAL HUMAN GAIT

The models reported in literature formulate human gait with a minimum of three segments and up to seventeen (and more) segments [4],[2]. The configuration of the seven segment skeleton in sagital plane is depicted in Figure 1. It is composed of the HAT and two legs, where each leg has a thigh, a shank, and a foot. By including the foot, the model expresses the important functions of absorbing and smoothing the ground impact, body balance during stance, and push off at late stance. The system possesses ten degrees of freedom in SLS and is derived by six torques τ_i $(i = 1, .., 6)$ around the joints and three forces which supply the translational accelerations at the stance foot. Its dynamics is defined by the following equations of motion,

$$M_{(q)} \frac{d^2q}{dt^2} + v_{(q,\dot{q})} + g_{(q)} = F_{(q)} \cdot \tau \qquad (1)$$

Where q and $\dot{q}$ are the $(n = 10)$ vectors of generalized coordinates and its time derivative, respectively, $M_{(q)}$ is a $(n \times n)$ mass and mass moment of inertia matrix, $v_{(q,\dot{q})}$ is the $(n \times 1)$ vector of generalized centrifugal and coriolis forces, $g_{(q)}$ is the (n) vector of the gravitational forces acting on the center of mass of each sub-body, F is the $(n \times 9)$ coefficient matrix of applied torques and forces, and τ is the (9×1) vector of moments and forces appleid at and on the joints.

The EoM's are implemented as inverse dynamics in MATLAB^{TM}. The measured kinematics (from the gait lab. of 'Otto Bock' Duderstadt in Germany) during SLS is first evaluated by a stick-figure animation. Then the simulation is performed by substituting the complete joint kinematics into the equations of motion (Eq. 1) and computing the ground reaction forces and joint moments of force.

The computed ground reaction force shows agreement with the properties of subject's weight, polarity change of the horizontal component after mid stance, acceptable ratio between the horizontal and vertical components, heel strike at weight acceptance, inertial body lift-off at mid stance, and the push-off effect of the foot-ankle mechanism at late stance. Therefore we conclude that the model represents the real key motions in a realistic way, and therefore we proceed with the assumption that the non-measurable internal joint moments of force, being determined by the simulation, will represent realistic values.

PROSTHETIC GAIT

Prosthetic gait is characterized by the fact that part of the lower extremities are replaced by artificial limbs. The important difference between the natural limbs and the artificial (prosthetic) limbs, according to the context of the present work, is that instead of muscles around the normal joints, the joints are passive, i.e. the knee and the ankle are characterized by a torque controlled by variable stiffness and damping. The motion of the prosthetic links is governed by the mechanical impedance of the artificial joint and by the amputee at the remaining (healthy or partially healthy) joints. For an A/K amputee, this is the hip joint and for an B/K amputee, these are the knee and hip joints. The other differences are the properties of mass, mass moment of inertia, single axis artificial joint instead of articulated surfaces (tibia-femur), and the lack of the wobbling effect of the flesh around the bones. The prosthetic gait model will be based on the normal gait model which was developed in Section I, with imposing the properties of the artificial limbs as described above. During swing, the dynamic model is of a double pendulum, and during stance, it is of an inverted double pendulum with moment acting at its both ends.

The purpose of this section is to compute the values of the stiffness and damping required at the prosthetic knee joint, to estimate the resulting prosthetic hip torque, and to predict the prosthetic trajectories. The results of this part will serve the next step of designing the controlled A/K prosthesis, its sensing and control. The knee mechanical impedance (KMI) is defined as the relation between the joint resistive torque and its relative angular velocity in the time domain,

$$Z = \frac{\tau_2}{\dot{q}_2 - \dot{q}_1} \tag{2}$$

The resistive knee torque τ_2 will be formulated as a linear combination of the joint state $\mathbf{x} = (\Delta \dot{q} \; \Delta q)^T$; $\quad \Delta q = q_2 - q_1$ with time dependant coefficients:

$$\tau_2 = -(k_0 + k)\Delta q - (d_0 + d)\Delta \dot{q};$$
$$k, k_0, d, d_0 \geq 0, \quad \Delta q \leq 0$$

k_0 and d_0 are constant bias values of stiffness and damping defined by aspects of realization. d_0 is due to the orifice area when the valve is fully opened, it can be minimized at the design stage of the hydraulic or pneumatic valves, but can not be reduced to zero. The relative angular position at the knee δq, is zero at full knee extension, and thus can not be negative. Furthermore, it is assumed that the natural coordination of the motions is made such that minimum energy is consumed and minimum effort is applied to produce minimum work in moving the body CoM from an initial location point to a next location point and from an initial segment state to a next state. Effort is here defined as the time integral of all absolute values (or square) of the joint torques. The numerical computation of the KMI which is a function of the instantaneous body state, can be computed using algorithms of nonlinear programing, and naturally involves the dynamic equations of motion. The problem is to find one 'best' hip torque history and the parameters of the KMI by defining the following optimization problem.

The Optimization Problem

The optimization problem can be stated as follows: given the experimental gait kinematics of a normal subject $\mathbf{q}^{ref}(t)$ and its first and second time derivatives, find the control torque vector $\mathbf{u}^*(\mathbf{t})$ that will generate the trajectories $\mathbf{q}(\mathbf{t})$ which will, as close as possible, track the normal trajectories $\mathbf{q}^{ref}(t)$ of the same subject, satisfying the underlying constraints in an optimal way.

The control torque vector $\mathbf{u}$ to be found is defined as:

$$\mathbf{u} = \; (\tau_1 \; d \; k)^T \tag{3}$$

with $\tau_1(t)$ the hip torque, d and k are the KMI parameters which minimize the scalar objective function defined as:

$$\min \; J = \sum_{i=1}^{N} \delta \mathbf{q}_i^T \delta \mathbf{q}_i + \delta \; \dot{\mathbf{q}}_i^T \; \delta \; \dot{\mathbf{q}}_i \tag{4}$$
$$\delta \mathbf{q}_i = \mathbf{q}_i^{ref} - \mathbf{q}_i$$

with respect to the state vector $\mathbf{x} = (\dot{\mathbf{q}} \quad \mathbf{q})^T$, and the control vector $\mathbf{u}$ which both satisfy the equality constraints of the equations of motion:

$$M_{i(q)} \frac{d^2 \mathbf{q}_i}{dt^2} + v_{i(q,\dot{q})} + g_{i(q)} - F_{i(q)} \tau_i = 0 \qquad (i = 1,..,N) \tag{5}$$

$$\begin{aligned} \tau_i \quad &= \quad (\tau_{1i} \quad \tau_{2i})^T \\ \tau_{2i} \quad &= \quad -(k_{0+}k_i)(q_{2i} - q_{1i}) - (d_{0+}d_i)(\dot{q}_{2i} - \dot{q}_{1i}) \end{aligned} \tag{6}$$

with the additional equality constraints of boundary conditions:

$$\delta \mathbf{q}_0 = 0, \quad \delta \; \dot{\mathbf{q}}_0 = 0, \quad \delta \mathbf{q}_N = 0, \quad \delta \; \dot{\mathbf{q}}_N = 0 \tag{7}$$

and the inequality constraints of bounds on d , k, and knee extension:

$$d_0, d_i \geq 0; \qquad k_0, k_i \geq 0; \quad q_{2i} - q_{1i} \leq 0 \tag{8}$$

The computation of the KMI during swing is made for medium walking speed of 1.58 $\left[\frac{m}{\sec}\right]$ and the results

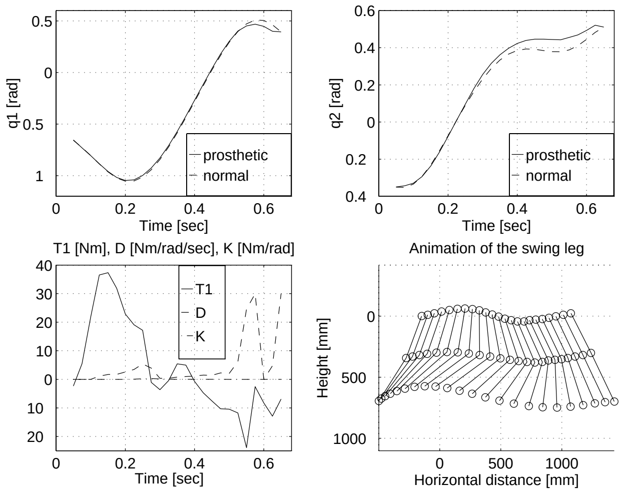

Figure 2: The KMI during swing

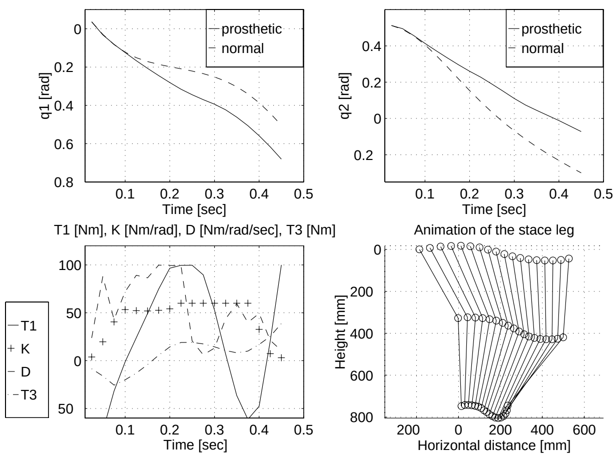

Figure 3: The KMI during stance

are presented in Figure 2. The optimum control vector includes both stiffness and damping and its initial values are taken as zero for the stiffness and damping, and normal value for the initial hip torque τ_1 .

The prosthesis seemed to well track the normal swing trajectories (a motion which is almost balistic) what can also be seen by the animation in Figure 2.

The KMI during stance is computed for the same walking speed. The optimization results are presented in Figure 3. The prosthesis tracks the normal trajectories with an error due to the load acting on the leg and the missing extension muscle moment. This can also be seen by the animation in Figure 3 by an almost constant knee flexion with no extension at late stance.

Figure 4: The laboratory prototype of a controlled A/K prosthesis used by a patient in the clinic

IMPLEMENTATION OF A CONTROLLED PROSTHESIS

The prosthesis is presented in Figure 4 and is composed of commercially available prosthetic foot and socket, a servo controlled hydraulic knee damper, a sensing system, computer and electronic system, and a mechanical construction. The design and component selection were directed towards a future product considering reliable functionality, low cost, low weight, low volume, and availability (catalogue items).

The instantaneous state of the prosthesis during walking will be continuously measured and the appropriate resistive torque will be applied at the knee. The measurements were selected to observe all possible prosthesis states during gait, sitting, standing, and slope/stair ascent/descent. The measurement (magnitude and direction) of the ground reaction forces during stance is significant for direct detection of weight bearing. The ground force vector can also signal the intention of the amputee and execution of stair ascent/descent. The knee relative angular position measures the knee flexion which is significant for controlling the hydraulic damper and can complete the identification of the prosthesis state. The computer, in addition to whole system management, also collects the measurements, it processes and generates the control commands and delivers these commands to the hydraulic damper at the knee.

Prosthesis Control

The literature reports on few documented controlled A/K prostheses and the few commercial prod-

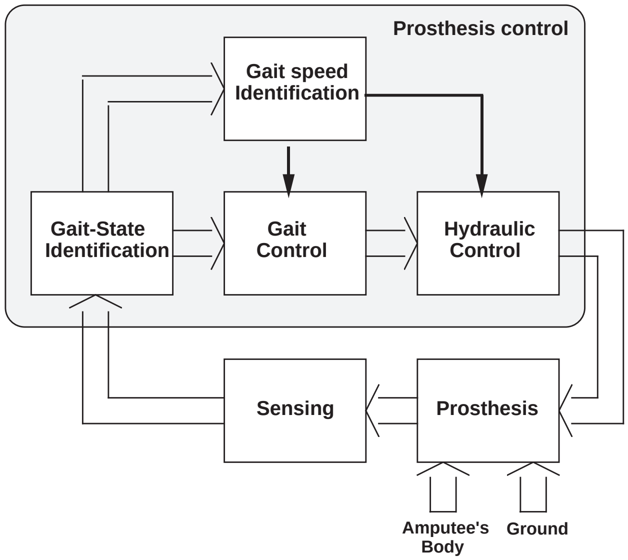

Figure 5: The block diagram of the prosthesis control system

ucts control the knee during part of the gait cycle [1], [3], and [8]. Interfacing with the nerve system [5] is still in research and is not yet known to function satisfactorily. In the case under consideration, a controller based on the locomotion dynamic model (even a linearized model) will probably be very complex, non robust and thus not implementible for real-time control. The controlled prosthesis will have special features, some not available by conventional prostheses, and it also has the capability to integrate and simultaneously perform them. These features include controlled knee flexion at early stance, high stability during weight bearing with a single axis knee, controlled knee release at late stance, controlled heel rise at early swing, damped full knee extension at late swing, damped knee at stance of stair descent and in sitting down, and adaptation to gait speed. The fact that the controlled knee is a single axis joint, makes the prosthesis more useful in different activities other than walking, like some light sport activities, riding a bicycle, or driving a motorbike.

In the following, the control concept which is implemented as a rule base system and composed of gait state identification, gait speed estimation, and knee control, will be described.

The selected control approach is based on the concept of 'soft' control (non-analytic) . The so called 'behaviour' of the system is governed by a finite set of rules. The pre-defined and stored parameters, models, and rules are combined as the knowledge-base of the system. This knowledge is based on existing experience and known facts about both normal and prosthetic gait. The present and past measured signals from the prosthesis will compose the data-base. The 'soft' control is algorithmic and is composed of four sub-functions described in Figure 5. These are gait state identification based on sensor measurements, gait control which generates the knee flexion and extension damping levels according to the present and past gait states, hydraulic control which generates the hydraulic valve angular position, and gait speed identification. The gait speed identification is based on measuring the time period between events. The gait speed level influences both the gait control and the hydraulic control.

In automation and logic systems, the concept of finite automata is also an attractive candidate. The aspect of man-machine interaction discussed in [6] calls for a 'natural' way of interaction rather than a complete autonomous control of the machine. The prosthesis control will be designed with a prior knowledge of the process input-output relations using human interpretation of both state identification and gait control by constructing a knowledge base and an hierarchical structure. The knowledge is represented by a set of rules (productions), the data-base contains the present and past sampled data stream, and the inference engine is executing the reasoning by scanning the rule-base searching for the rule that is satisfied, then it coordinates and executes the sequence of actions resulting from the rule that was obeyed (fired). In constructing the knowledge base of the system, two types of experts are involved, the biomechanical engineer and the clinician. The biomechanical engineer is familiar with the technical structure and functionality of the system and he is involved in constructing the knowledge base of the state identification and hydraulic knee control. The state identification relates the physical measurements to the heuristic gait states which are known to the clinical expert. The clinical expert then constructs the knowledge base of the prosthetic-gait control which relates the instantaneous heuristic gait state to the proper prosthesis response. Both experts are involved in the third phase of modifications during clinical tests.

The periodical gait characteristics are described by the block diagram in Figure 6. One gait cycle is divided into two main periods which are the stance, when the prosthesis accepts the body weight and supports it, and swing, when the prosthesis is advanced forward. The important events (short duration) and their resulting states are heel contact

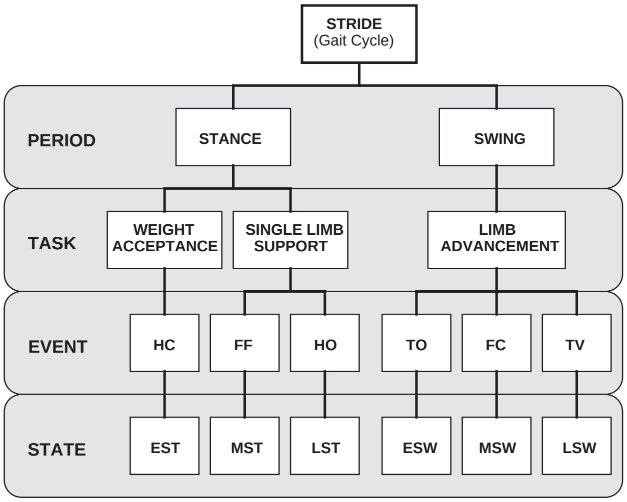

Figure 6: The periodical gait chracteristics within one stride

(HC) at early stance (EST), then foot flat (FF) at mid stance (MST), and heel off (HO) at late stance (LST). The swing period starts with toes off (TO) at early swing (ESW), then foot clearance (FC) with the ground at mid swing (MSW), and ending with tibia vertical (TV) at late swing (LSW) until full knee extension towards next HC.

Gait State Identification

For the gait state identification, the inputs which trigger state transitions are the observable short duration events defined in Figure 6. The events are detected by observing when each of the measurements crosses a preset threshold level and enters into one of the discrete (level) ranges. The collection of the discrete ranges for each one of the three measurements (as members) are combined into finite sets, using alphabet of symbols as:

$$\begin{aligned} Threshold(f_t) &\in (Positive, Negative) \\ Threshold(f_r) &\in (Low, Mid, High) \\ Threshold(dq) &\in (Low, High) \end{aligned}$$

where $Threshold(.)$ is a magnitude segmentation function and the three selected measurements fr, ft, and δq, are respectively the radial and tangential force components, and the knee relative angular position.

The gait states are the periods which are initiated by the events and the finite set of actions is a collection of discrete extension and flexion pairs (levels) of knee damping. The finite set of input names (events) to be detected is defined as:

$$Input \in (Heel\_Contact, Foot\_Flat, Heel\_Off, ... \\ ..., Toe\_Off, Foot\_Clearance, Tibia\_Vertical)$$

The finite set of states initiated by the events is defined as:

$$State \in \\ (Early\_Stance, Mid\_Stance, Late\_Stance, ... \\ ..., Early\_Swing, Mid\_Swing, Late\_Swing)$$

Hydraulic Control

The control rule-base relates the extension and flexion damping to the different identified gait states. The finite set of actions to be generated (for a single valve hydraulics) is defined as:

$$Extension\_Flexion \in \\ (Null\_Null, Null\_Low, Null\_High, Null\_Full, ... \\ ..., Low\_Full, High\_Full, Full\_Full, Full\_High... \\ ..., Full\_Low, Full\_Null, High\_Null, Low\_Null)$$

Gait Speed Detection and Application

The gait speed is changing from one gait cycle to the next and also within the cycle. It is required that the gait speed estimation will be invariant to gait characteristics, such as forces and angular position profiles, because these will be different for different subjects. It is also required that the speed will be updated within one cycle and that the knee will react immediately after the change in speed occurs. The speed is inversely proportional to the time period from a gait event in one gait cycle to the same event in the next cycle. Six counters are counting the 'time' between events. Since at least six events are detected during one cycle, it is possible to have six speed updates per cycle. The average speed at time i is computed as a weighted sum of counts from the last three state transitions.

Sequential Finite-State Control (SFSC)

Finite automata is one of the concepts for implementing non-analytical control of a dynamic system. The application of finite automata to artificial motor control is offered in [7]. Its properties of robustness to measurement noise, reliability in addition to natural and relatively easy implementation, and adaptation, make it attractive for the case under consideration.

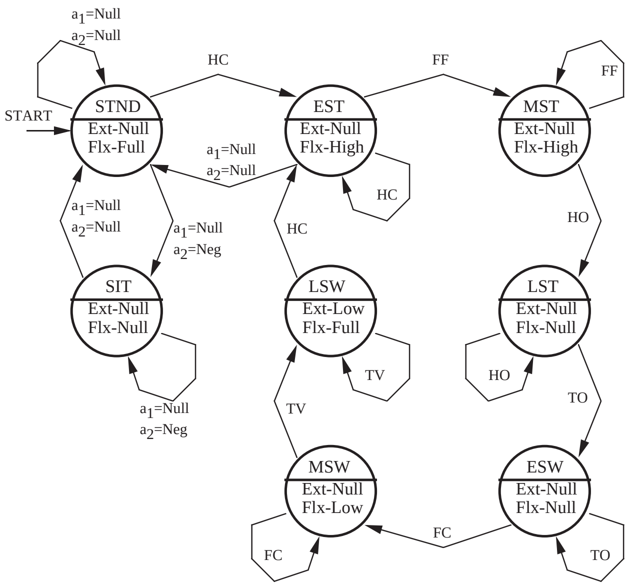

Figure 7: The digraph of the sequential finite state control

Gait is a continuous body motion generated by a sequential process of repeated motion patterns. The sequential process is repeated in each gait cycle and can be divided into a finite number of states. The sequential finite-state machine is a process which is defined by a transition function and an action function. The transition function f_s is defined as follows:

$$S_{i+1} = f_s(S_i, I_i) \tag{9}$$

where S_i, S_{i+1} are the present and next states respectively and I_i is the present input. The action function f_a is defined as:

$$A_i = f_a(S_i) \tag{10}$$

where A_i is the resulting output action made at the entrance to state S_i.

The segmentation of the measured signals and the identification of the gait events are made first. The transition to the next state is generated via a transition function depending on the present event and the last gait state. The flexion/extension damping is determined via the rule-base of the inverse hydraulic model.

The clinical gait tests with the controlled prosthesis were made by an A/K amputee (age of 60's, and weights of about 80 Kg) who is normally using a simple wooden prosthesis with a 'friction' articulated knee joint and SACH foot. About 30 repeated gait trials were successfully made. The identification and control

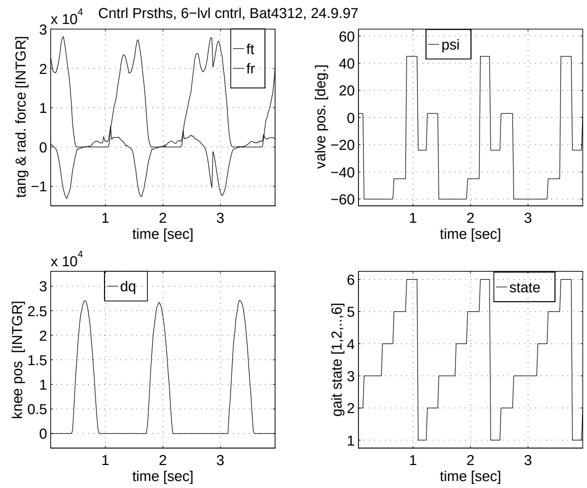

Figure 8: The performance of the SFSC. Measured variables (left), state identification and valve command (right)

were performing well and both the amputee and the orthopedist were satisfied with the performance. The amputee reported on a relatively fast adaptation period to the controlled prosthesis since it immediately transmits a safety feeling by responding properly to body load. In addition, swinging the prosthesis forwards was easy and the damped full extension at late swing was easily (well) adjusted. The orthopedist, by observing the amputee gait, was reporting that the ease of usage is shown by the relatively low effort required from the amputee by the way he combines the rhythm of his whole body during gait.

The performance of the SFSC is presented in Figure 8. The figure shows the measured variables, the gait state identification, and the angular position (control command) of the hydraulic valve. It can be noticed that the controller performs smooth transitions from one state to the next state and showed robustness to faulty inputs.

SUMMARY AND CONCLUSIONS

The research and development work on the controlled A/K prosthesis was divided into three parts where each part deals with a different problem but each problem is supported by the preceding part combining the whole work with a logical sequence and methodology.

In the first part the kinematics and dynamics of human gait were studied and evaluated. The process of using the software package of AUTOLEV^{TM} for generating the equations of motion was found very efficient. It is consuming relatively short time, its modularity enabled easy modifications and upgrading, and having the complete symbolic terms, contributes in giving a good insight from the physical point of view. The implementation of the equations of motion for simulation in MATLAB^{TM} was also found efficient, easy to program with a self explanatory language, and with expressive graphic capabilities. The important contribution of Section I is in presenting an alternative, simple configuration, for modeling the foot-ankle mechanism. A detailed and correct modeling of the foot-ankle mechanism was proved to be valuable for analyzing prosthetic gait and can be further used for optimizing feet and below knee prostheses.

The second part presented a new method to compute the optimum mechanical impedance of a prosthetic joint, the predicted performance of the controlled prosthesis, and the moment input (effort) by the amputee. Although the presented work was not intended to accurately compute the optimal mechanical impedance of a prosthetic knee due to the simplifying assumptions taken, the results are reasonable and can also contribute to the design of a passive B/K and A/K prostheses.

The third part presented a non analytical and clinically tested control concept and its performance. The prototype prosthesis with its hydraulics, electronics and sensing, performed well in all (about six) clinical sessions. The sensing, identification, and control repeated their performance from session to session although the prosthesis is not always attached to the stump in the same manner i.e. orientation and depth. The contribution of Part III is in presenting a systematic procedure of developing a knowledge base controller starting with a simple finite state control towards combining it with a fuzzy logic controller. An effective way for estimating the gait speed as an important gait determinant was also presented. The existing prosthesis is a good basis towards further improvements, clinical tests, and a commercial product design.

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