A MICROCOMPUTER BASED KNEE-LOCKING CONTROL SYSTEM

M. Cvetinović
V. Cvetković

In recent years more attention has been paid in designing improved assistive devices using new technologies. It appears that micorprocessor controllers have found applications in prosthetics, too. An improved understanding of the problem should precede the study of the general controller concept since adequate control implies recognition of the force and position feedback from the prosthesis. None of the existing devices fulfills generally stated problem. Current work must be regarded as preliminary, and the mechanically and hydraulically controlled prostheses will continue to be used. The only question is not whether the microprocessor is necessary for decision and control of an above - knee prosthesis, but from what (or how) should the controlling be done.

One approach is to use a knee-locking system controlled by electromyographic (EMG) potentials from remanent muscles in the amputee's stump [1]. The second one promising control concept is to imitate the gait pattern of the normal leg [2]. Another possible way for knee-locking control is to use position sensors and foot switches. In an early work [3], the application of finite state theory based on these sensors was proposed. However, technology did not permit the practical realization of this concept at that time.

Few reasons could be considered for experimental study of microprocessor controlled knee-locking system:

- Security form fall; microprocessor controller is capable to detect some abnormal situations and takes appropriate action. It ought to be pointed out that this property of the system should have some important psychological effects.
- Improved getting up sequence; it enables full support during getting up regardless whether the prosthesis is completly extended or not. This also should be done for safety reasons.
- Damping coefficient modification; it is worthwhile to experiment with damping coefficient modification if it can be done with particular prosthesis.
- Improved knee-lock phasing; conventional prostheses lock in hyperextension, but some of them (using appropriate controller) can be made,

to a certain extent, to follow extend-flex-extend pattern similar to natural limb. It is expected that vertical displacement of the body centre of gravity and accompaning energy requirements for the amputee are less than with conventional system, after the patient became accustomed to the prosthesis.

Control Algorithm

One way to take into account the above mentioned considerations, is presented in global algorithm sketched in fig.1. It is assumed that foot switches and vertical position and knee angle transducers are available inputs.

Logically, the control sequence can be devided into four subunits: standing, flexion and extension make normal gait loop while the sitting and getting up part is the fourth macro state.

The first test should be whether the patient is in sitting position. If yes, the sitting and getting up control sequence is entered. On no, standing phase test is performed. Until conditions for standing are violated the test is repeated. Then, flexion state starts. If necessary, speed regulation may be included. After appropriate boundary value is reached, the intention for sitting down test is performed. If yes, sitting and getting up routine is entered. Here again, the suitable damping coefficient adaptation may be introduced. Next phase is extension. Few points should be mentioned; swing phase damping adjustment, delayed locking which enables controlled felxion. This results in smaller instantaneous joint loads at the beginning of stance phase. By monitoring knee angle, hitting an obstacle or similar hazard situation can be detected and appropriate action undertaken (e.g. some damping should be allowed instead of momentary locking). After full support has been gained the main loop is restarted.

Experimental Controller

In order to test realizability of the above algorithm, a simplified version has been implemented using 8080 microprocessor around modified Dynaplex hydraulic gait control unit. The following sensors have been used:

A - heel contact switch
B - middle of the sole contact switch
C - tiptoe contact switch
α - knee angle transducer (fig.2.)
V - vertical position transducer (fig.2.)

For fracking α and V angles 8 bit A/D converter has been used.

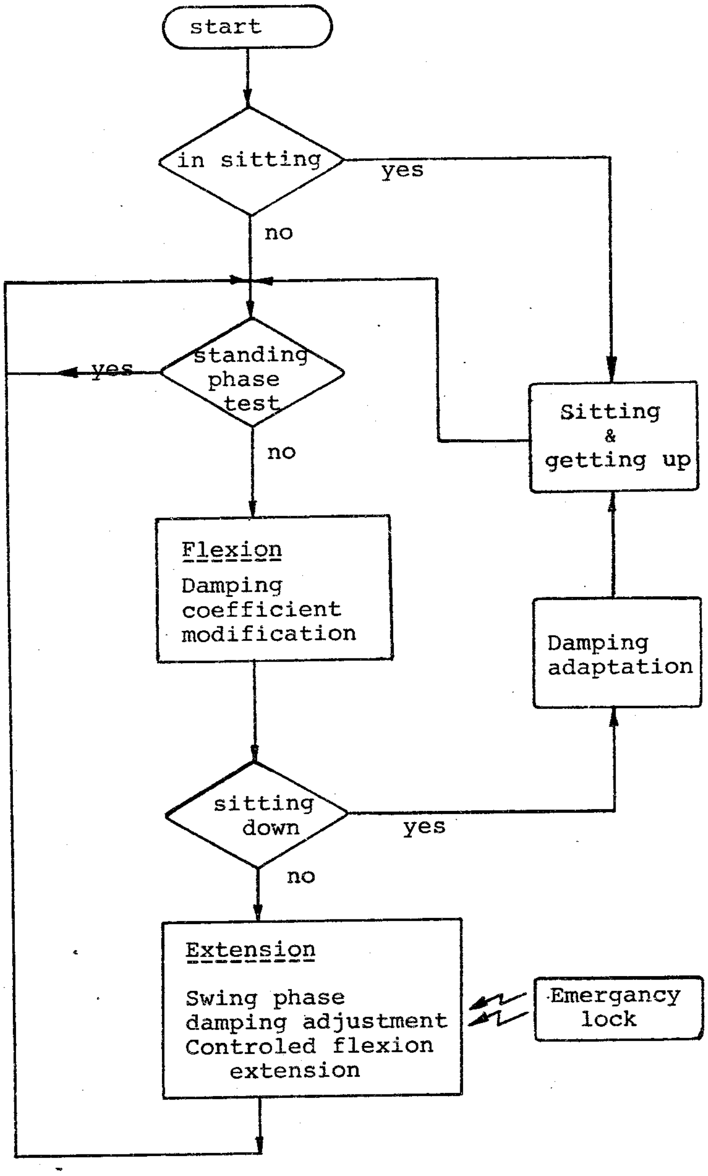
start
in sitting
yes
no
standing phase test
yes
no
Sitting & getting up
Flexion
Damping coefficient modification
Damping adaptation
sitting down
yes
no
Extension
Swing phase damping adjustment Controled flexion extension
Emergancy lock

Fig. 1.

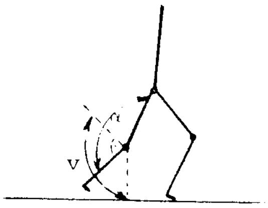

Fig. 2.

Detailed block diagram for microcomputer controlled unidirectional knee-locking device is presented in fig. 3(a,b). The flow of the program is straight-forward and only few explanations are necessary. Before in sitting position test, A/D conversions of α and V angles are performed. This test is expressed as inequality $\alpha + V < K_1$. Boundary volue K_1 is assumed to be about 190°. As seen in fig.4.a, it thigh is in horizontal position, $\alpha + \beta = 180^\circ$. The angle s (about 10°) defines the region of tolerance. Therefore, obviously, $K_1 = 180^\circ + s$

(a) (b)

Fig. 4.

Intention for gait is detected when heel contact (switch A) is turned off. Flexion loop is repeated until tiptoe contact exists, or until the instantaneous value of α becomes less than α_1. A reasonable value for full flexion is assumed to be about 120°, but it can be adjusted to the patient's needs on the spot.

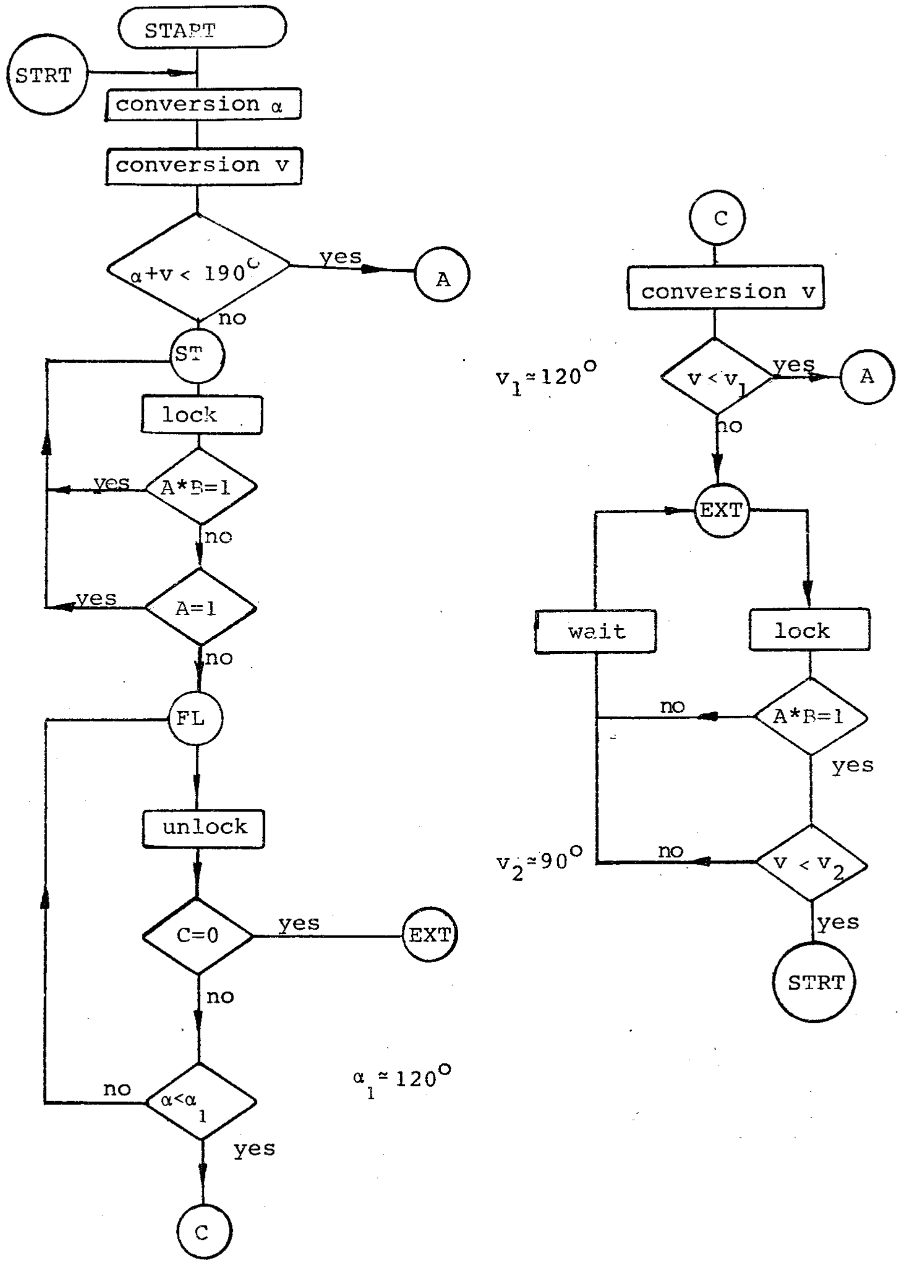
START
STRT
conversion α
conversion v
α+v < 190°
yes
A
no
ST
lock
A*B=1
yes
no
A=1
yes
no
FL
unlock
C=0
yes
EXT
no
α<α1
no
yes
C
α1 ≈ 120°
C
conversion v
v1 ≈ 120°
v < v1
yes
A
no
EXT
wait
lock
A*B=1
no
yes
v2 ≈ 90°
v < v2
no
yes
STRT

Fig. 3-a

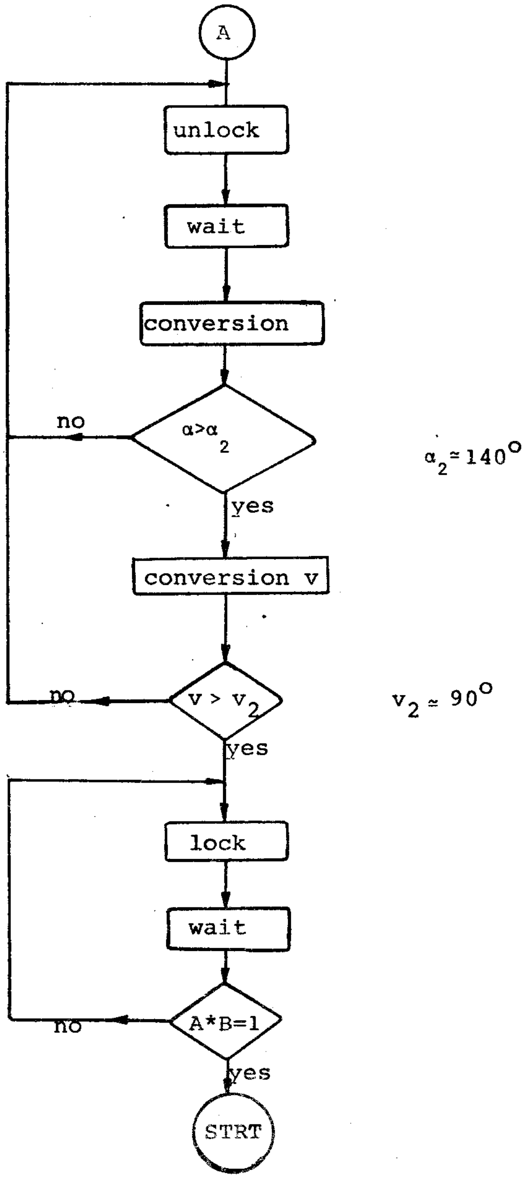
A
unlock
wait
conversion
α>α2
no
yes
α2≈140°
conversion v
v > v2
no
yes
v2 ≈ 90°
lock
wait
A*B=1
no
yes
STRT

Fig. 3-b

At this point, the test for sitting down is performed. The meaning of the constraint V_1 is explained in Fig. 4.b. Angle s is again in the region of tolerance. If $V < V_1$, then the sitting down and getting up routine is entered. If not, or in the case when tiptoe contact does not exist, the device goes over to the locked state.

After the former test has been successfully completed, the extension phase begins. At this stage, no attempt has been made to control damping coefficients which are already built in unidirectional dyna-flex hydraulic gait control unit. Therefore, the flexion is prevented until the shank is again in vertical position.

The sitting down and getting up routine is presented in Fig.3c. Point A is entered when, after the initialization, the sitting position has been detected. The device stays in unlocked state during sitting and rising until $\alpha > \alpha_2$ and $V > V_2$. Having reached those values, the device is forced into the locked state. In this way, the patient should feel safe from falling down while rising. The device stays in this state until full support is gained.

Conclusion

This paper presents a knee locking control algorithm implemented in the self-contained single board micorcomputer. Experiments were successfully carried on in simulated conditions [4]. Foot switches, knee and vertical position transducers were sensed and taking into account the previous state, control signal was generated. This, externally controlled device, in contrast to the EMG controlled system, is expected to be more reliable because of its digital nature. However, the latter has inherent advantage of instantaneous detection of "intentio s" for changing state and possibility for proportional control. Perhaps, the combination of two should be investigated. Alternatively, further improvement could be obtained by using pressure transducers instead of simple switches. However, before thinking of more sophisticated controller, complete clinical evaluation of the existing one is necessary. It should be pointed out that those tests will be performed with technologically advanced version which is currently under development.

References

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