THE CONSTRUCTION AND EVALUATION OF A LOW-COST ISOKINETIC KNEE EXERCISER

David Greenblatt*

Summary

An inexpensive knee exerciser which closely approximates the performance of state-of-the-art isokinetic equipment has been constructed and a preliminary evaluation has been carried out. Unlike other exercisers, the lower limb angular velocity cannot be set *a priori* but, on the basis of patient performance it is possible to restrict limb velocity to within upper and lower bounds. The exerciser makes use of a shock absorber and lever mechanism which allow easy calibration of the system while elementary electronics and computer interfacing allow the measurement angle, velocity and torque in real time. A significant advantage of this exerciser over some isokinetic knee exercisers currently on the market is that limited arc rotation can be enforced. This type of limb restriction is important in many rehabilitation programs.

Opsomming

'n Lae koste knie oefenmasjien wat die werkverrigting van huidige vooraanstaande isokinetiese masjienerie nou naboots was gekonstrueer, en voorlopige toetse is gedoen. Anders as ander masjiene, kan die rotasie snelheid van die onderste ledemaat nie *a priori* verstel word nie, maar 'n minimum en maksimum perk kan gehandhaaf word soos bepaal deur die pasiënt se krag. Die oefenmasjien maak gebruik van 'n skokbreker en hefboom meganisme wat kalibrasie vergemaklik. Eenvoudige elektroniese en rekenaar koppeling monitor die hoek, snelheid en wringkrag terwyl die masjien in gebruik is. Hierdie masjien het 'n aansienlike voordeel bo vele ander isokinetiese knie oefenmasjiene in gebruik omdat die rotasie hoek van die ledemaat beperk kan word soos verlang. Dié beperking van die ledemaat is belangrik vir baie rehabilitasie programme.

Key words: low-cost, isokinetic knee exerciser, shock absorber/lever mechanism, computer interfacing, limited arc rotation.

INTRODUCTION

Ever since isokinetic exercise was introduced by Thistle *et al*[1] it has been widely used for research, clinical testing and rehabilitation, and has become an indispensable tool in the field of sports science. The main advantage of this type of exercise over other modes is that it allows the development of maximal muscle tension throughout the range of limb motion (see Thistle *et al*[1] and Hislop and Perrine[2]). Exercisers of this kind currently on the market include the highly sophisticated electro-mechanical Cybex system which is capable of exercising many body joints, while hydraulics-based systems such as the Orthotron KT series are used exclusively for knee exercise and rehabilitation.

For many years, South African physiotherapists have been aware of the important and essential role that isokinetic exercise plays in physical rehabilitation and injury assessment. It is unfortunate, however, that the cost of importing and maintaining isokinetic testing and rehabilitation equipment is prohibitive, and this limits its availability in the country. During late 1990 it came to the attention of South African physiotherapists that the Bioenergetics of Exercise Research Unit (BERU) at the University of Cape Town (UCT) are "considering the development of an isokinetic unit which will be calibratable, give reproducible results over the entire range of movement and will measure torque and angle simultaneously in real time".

An inexpensive knee exerciser of this type has recently been developed privately and has been tested at the University of the Witwatersrand Mechanical Engineering Laboratories. It will be shown in the forthcoming sections of this paper that the exerciser closely approximates the isokinetic principle for a fraction of the cost of overseas exercisers and fulfills all of the requirements set down by the BERU.

PRINCIPLE OF OPERATION

An ideal isokinetic exerciser will allow the patient's limb to move at a constant velocity irrespective of the torque developed. This is achieved, for example, with the Orthotron by means of pressure compensating values. Another less obvious method of achieving isokinetic motion is to determine *a priori* the patient's torque versus angle characteristics and then set up the system response so as to bring about isokinetic motion.[3] The problem of designing such a system is greatly simplified if it is assumed that the torque versus angle characteristics for all patients can be characterised by a certain family of curves. By considering the results from a large cross-section of patients, it is evident that for both extensor and flexor muscles of the knee, torque is at a minimum at the fully extended and flexed positions and reaches a maximum somewhere in-between. This type of result is shown in figure 2 with reference to the nomenclature of figure 1.

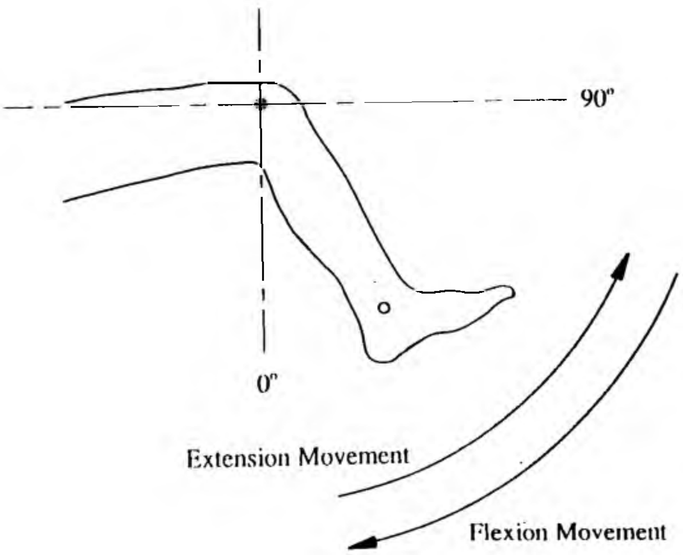

Fig 1: Nomenclature for the knee position during isokinetic exercise

The exerciser that is described here may be configured to match the torque versus angle characteristics of the patient in one direction, and consequently bring about isokinetic motion of the limb in either extension or flexion. This may also be achieved to an acceptable degree of approximation for exercise in both directions (see section 4). Figure 3 is a schematic representation of the exerciser mechanism where OB is the lever arm, AC is the position of an hydraulic shock absorber and the points O and C are stationary. The lower leg of the

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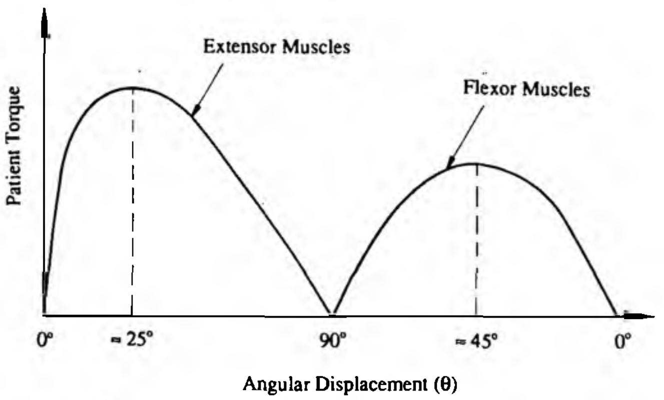

Fig 2: Typical torque versus angle characteristics for the knee joint

patient applies a torque T_P to the lever arm. Now the relative position of C to O can be adjusted so as to locate the position of maximum torque developed by the patient at $\phi=90°$. In so doing the patient is required to move the shock absorber through its maximum displacement relative to the limb motion in the region of maximum torque. It is an easy matter to show, and indeed intuitive, that the shock absorber resists with maximum force at this point when the motion is isokinetic. The magnitude of the resisting force, relative to the force of the patient, can then be set by suitably adjusting points A and C or by adjusting the damping capacity of the shock absorber. In this manner the patient's lower leg can be varied through a spectrum of velocities.

CONSTRUCTION AND MEASUREMENT

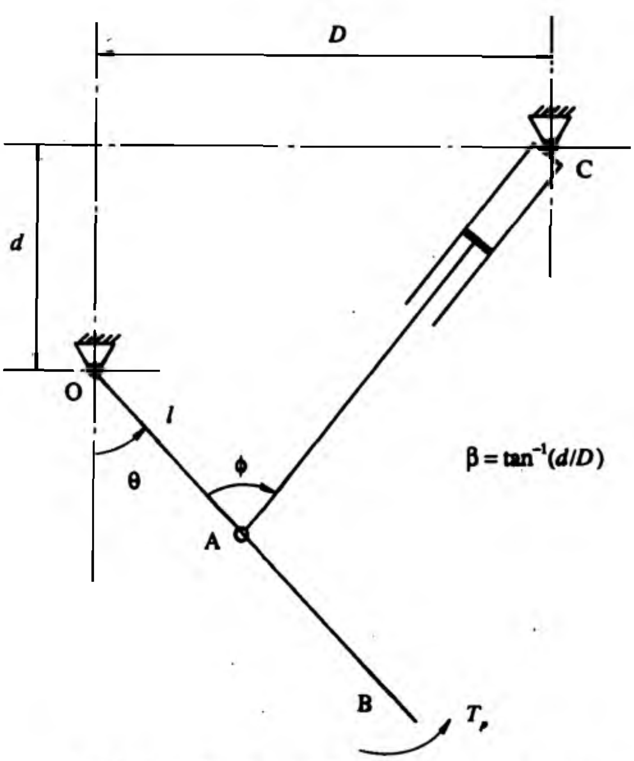

Fig 3: Schematic representation of the proposed exerciser

The exerciser constructed for this study made use of a chair similar to that used for the Orthotron. The Orthotron hydraulic assembly, however, was replaced by a shock absorber and lever assembly such as that shown in figure 3. The shock absorber had three settings corresponding to three different damping coefficients and the positions A and C were adjustable relative to O. This allowed the patient to traverse the entire range of lower leg motion and, in addition, arbitrary limited arc rotation could be enforced by suitable adjustment of positions A and C. It is important to note that this latter feature, which is often required in routine rehabilitation programs, is not available on the Orthotron KT series. The mechanism itself occupied a little less volume than the KT2 hydraulic actuator and a patient could easily vary the settings while remaining comfortably seated. Consequently, no bulkiness problems arose.

For the angular displacement (or angle, θ) versus time measurements conducted on the exerciser potentiometers, connected in wheatstone bridge configurations, were fitted to the left and right leg shafts at O. These were then interfaced with an IBM computer via a computer scope. A calibration of the experimental setup showed a linear response of bridge voltage output as a function of θ.

The differential equation governing the motion of the exerciser was derived by da Silva and Greenblatt.[3] For this analysis a substantially simplified approach was adopted in which it was assumed *a priori* that the motion of the exerciser was isokinetic. This allowed considerable simplification of the governing equation giving rise to a quasi-isokinetic analysis. Using this analysis the torque developed by the patient can be expressed as

$$T_p(\theta) = \frac{\omega c\, l^2\sqrt{D^2+d^2}\cos(\theta-\beta)\sin\phi}{\left(l^2+D^2+d^2-2l\sqrt{D^2+d^2}\sin(\theta-\beta)\right)^{0.5}},$$

where ω is the angular velocity of the lever arm and c is the linear damping coefficient of the shock absorber. The other quantities are defined in figure 3. Details of this derivation are given by da Silva and Greenblatt.[3] For calibration purposes the damping coefficients of the shock absorbers were determined for both extension and compression strokes by a simple "load and time" technique. Although this system is fairly complex geometrically, it is pointed out here that the dynamics of the system are not difficult to characterise mathematically. Consequently, with the geometry of the system and the damping coefficient of the shock absorber known, it was not a difficult task to calibrate the system.

RESULTS AND EVALUATION

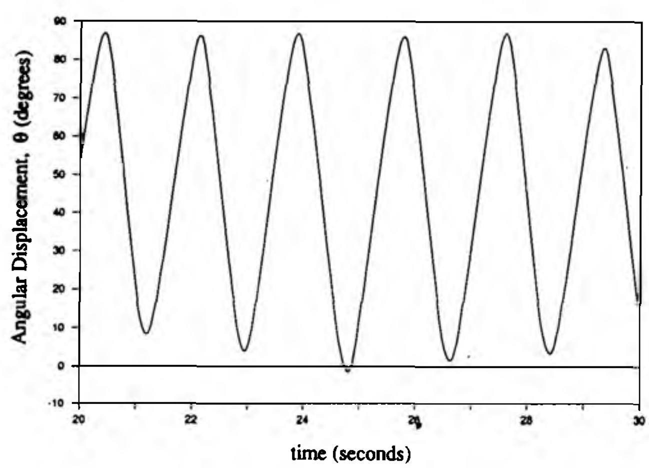

Fig 4: Angular displacement versus time plot for patient A – right leg

The exerciser was evaluated using a female "patient" and a male "patient" who are referred to henceforth as patient A and patient B respectively. Both patients were in their early twenties when this evaluation was carried out. In all tests performed, both legs were exercised simultaneously, with one leg flexed and the other extended and vice versa, as this proved to be just as convenient as exercising one leg at a time.

An example of the angular displacement versus time plot for the right leg of patient A is presented in figure 4. The left leg result is almost identical and consequently is not presented here. In order to achieve the result presented in the figure, the geometry of the exerciser was configured, on a trial and error basis, to ensure iso-

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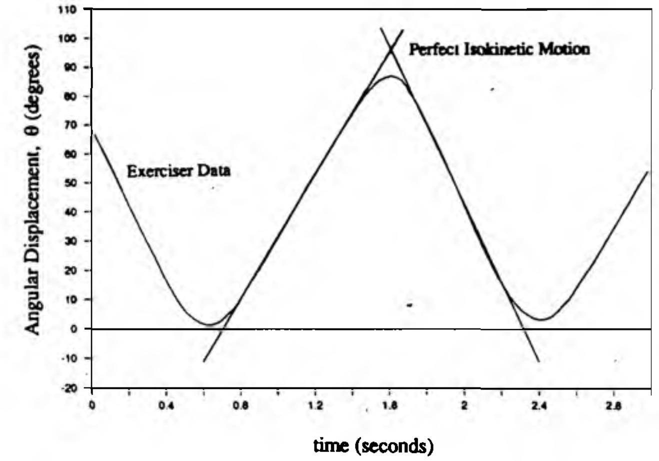

Fig 5: Angular displacement versus time plot for patient A – right leg, one cycle

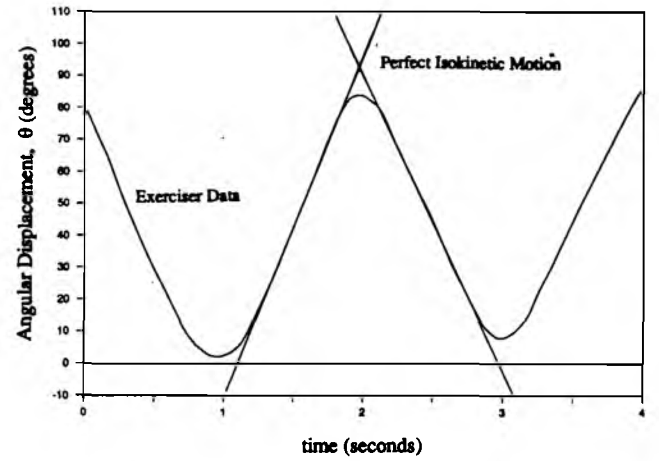

Fig 6: Angular displacement versus time plot for patient B – left leg, one cycle

kinetic motion for both extension and flexion. This is more clearly illustrated in figure 5 where only one extension and one flexion are considered. In this figure, linear curves have been superimposed on the exerciser data for both extension and flexion. These linear curves represent perfect isokinetic motion. It can be seen that for almost the entire range of lower leg motion, deviations from the isokinetic ideal are negligible. It is only at the extremities i.e. near 0 to 90 degrees, that significant departures from isokinetic motion are evident. These deviations are to be expected since it is impossible to physically achieve perfect isokinetic motion, as this would require infinite accelerations of the limbs at the extremities. Data for the left leg of patient B, in the same format as figure 5, is presented in figure 6. Once again isokinetic motion was achieved for both flexion and extension by suitable adjustment of the exerciser geometry. It should be noted, however, that the shock absorber/lever arm length OA, and consequently the mechanical advantage, was set to be the same for both patients.

Table 1: Angular velocities for both patients

	Extension (degrees/s)	Flexion (degrees/s)	Average (degrees/s)
Patient A	107	135	121
Patient B	107	92	99.5

Table 1 shows the lower leg extension and flexion angular velocities (ω) for both patients. These velocities were calculated from the linear slopes in figures 5 and 6. It is evident from the table that, for both patients, the velocities in the two directions are not the same. In order to achieve the same velocity in both directions the damping capacity of the shock absorber would have to change depending on the direction of motion. This can be achieved but was not investigated here since the exerciser was constructed merely to illustrate the concept. Nevertheless, it should be noted that even in the worst case presented here, i.e. patient A, the variation in velocity about the average is only 12%. For the rehabilitation of many conditions this may be a perfectly acceptable tolerance.

It is of fundamental importance to note the subtle difference between this exerciser and current state-of-the-art isokinetic equipment. Here, the patient's lower leg velocity is determined by the exerciser settings *as well as the patient's strength*. Consequently, in assessing a patient, the therapist must first ascertain what settings will give a particular velocity range and then use this information to advance the patient through a spectrum of velocities by lengthening or shortening the shock absorber/lever arm length OA. Naturally, the strength of the patient will increase with repeated treatments and so too then will the velocity at a particular setting. This feature would have to be accounted for by the therapist. When a patient uses the exerciser with arbitrary settings, angular velocity will vary with angular displacement. However, it is up to the therapist to configure the exerciser so as to ensure isokinetic or nearly isokinetic motion. Depending on the skill of the therapist the variation in velocity can be minimized and possibly eliminated.

If both legs are exercised simultaneously and the mechanism settings are the same, problems may arise with this exerciser, particularly if one leg is significantly stronger than the other. Under these circumstances the stronger leg will tend to move faster than the weaker leg. This would then lead to coordination problems as the left and right legs would become increasingly out of sync. It is important to note, however, that the left and right mechanisms are independent of each other. Therefore, the mechanisms could be adjusted independently with different settings so as to ensure the same average velocity for both legs. The results presented in the article show that, for a person with approximately equal strength legs, both legs can be exercised simultaneously, with both left and right mechanism settings the same, and the legs remain in sync.

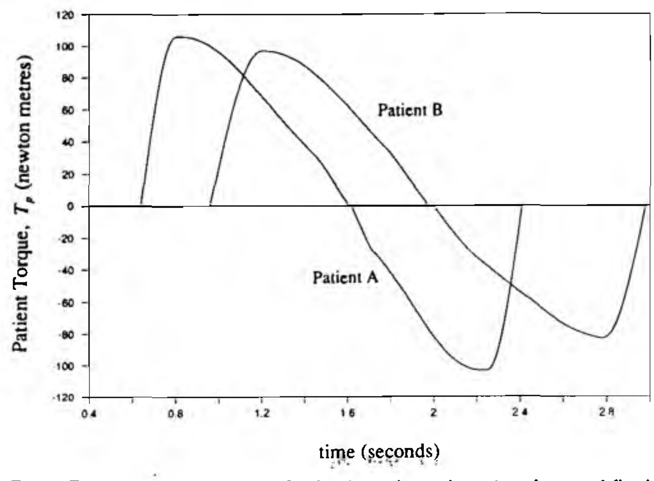

Fig 7: Torque characteristics for both patients in extension and flexion

The angular displacement and velocity results described above were used in conjunction with the equation presented in the previous section to obtain the torque curves for both patients. The results are presented in figure 7 where positive torque denotes extension while negative torque denotes flexion. For extension, patient A produced a maximum torque of 105 newton meters (Nm), while patient B produced 97 Nm. Even though their angular velocities were identical the right hand side shock absorber damping coefficient was slightly larger than that on the left hand side and consequently patient A had to develop a larger torque to attain the same velocity. In flexion, the higher velocity of patient A (see table 1) translated directly into a larger torque. This linear relationship can be clearly seen from the equation in the previous section. Here, patient A developed 103 Nm

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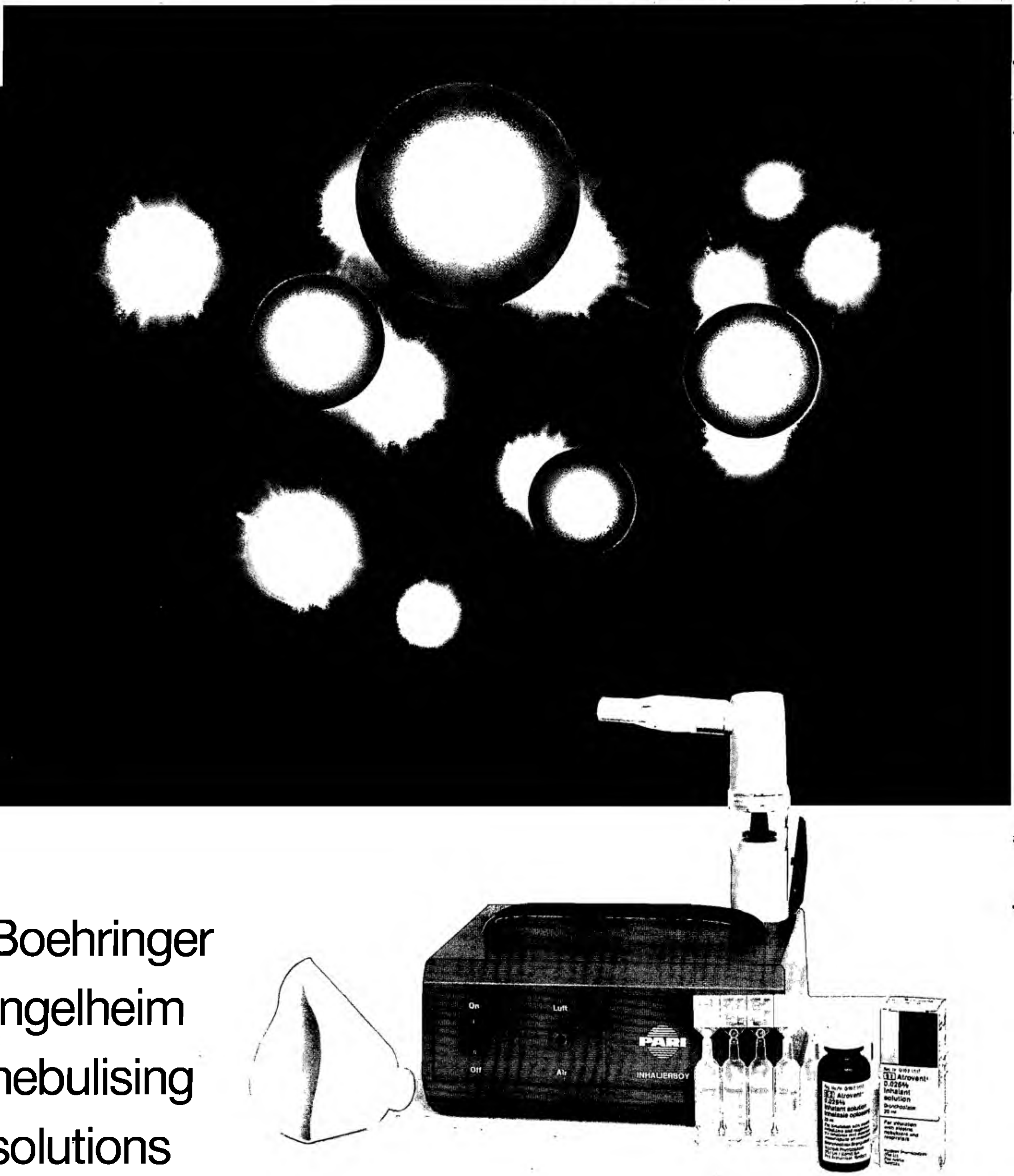

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whereas patient B could only achieve 83 Nm.

It has been shown in this section that the therapist must play a far greater role during rehabilitation than is the case when using other isokinetic equipment. This probably has additional merits of its own but is a vital ingredient to ensure the success of the exerciser. Indeed, it is a small price to pay when one considers that this exerciser will retail at a fraction of the cost of imported equipment. The dire need for affordable equipment in this country cannot be over-stressed.

CONCLUDING REMARKS

An inexpensive exerciser has been described here which closely approximates the performance of state-of-the-art isokinetic exercisers. Unlike other exercisers of this type, lower limb velocity cannot be set at the outset but the therapist can ensure a particular velocity range on the basis of patient strength. It is also evident from the foregoing sections that the two most expensive items on the exerciser are commercially obtainable shock absorbers. It is thus obvious that this exerciser will be dramatically cheaper than any of the isokinetic exercisers currently on the market.

In fulfillment of the BERU requirements it is evident that, provided the geometry of the mechanism and the damping coefficient of the shock absorber are known, the system is fully calibratable and is capable of providing reproducible results over the entire range of movement. It has also been shown in the above sections that simple electronics and computer interfacing allow the simultaneous measurement of torque and angle in real time. One definite advantage of this exerciser over those like the Orthotron is that it allows enforcement of limited arc rotation which is important in many rehabilitation programs.

ACKNOWLEDGEMENTS

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