Rehabilitation engineering

# Computer-controlled wheelchair ergometer

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**Abstract**—*A new wheelchair ergometer has been designed in which a combination of realistic simulation of wheelchair propulsion—with adjustable parameters for rolling resistance, air drag, wind speed and slope—and force measurement has been realised. The static solution enables the measurement of physiological and kinesiological parameters. All data from force transducers in seat and backrest, torque transducers in the wheels and force transducers in the wheelframes as well as the acquired speed are sampled in a data-acquisition system. An offline curve processor allows the acquired data to be processed with standard or custom-programmed routines. Preliminary results have been added and are discussed.*

**Keywords**—*Dynamometry, Ergometer, Propulsion, Wheelchair*

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## 1 Introduction

STUDIES ON wheelchair propulsion have mainly been focused on physical stress or on technical improvements. Some studies have tried to combine physiological and more biomechanical techniques to find criteria for optimisation of wheelchair design. In this analysis of the wheelchair/user interface, physiological and biomechanical measurements under known, manipulatable conditions which simulate real-life wheelchair propulsion, are needed. Measurements should not only include exact information on the applied workload, but also information on the forces applied by the user to overcome this workload and on the metabolic cost of doing so.

The validity of any parameters to be measured is directly related to the verisimilitude of the simulation. An ergometer must not only allow accurate measurement of power output and applied forces, but must also simulate wheelchair driving, at least mechanically, as realistically as possible. Moreover, because comparison of different wheelchair dimensions is required, the ergometer should allow easy adaptation of the wheelchair dimensions. A stationary ergometer is preferable, because it would facilitate the measurement of physiological and kinesiological parameters.

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The demands with which such an ergometer must comply can be summarised as follows:

(*a*) adequate simulation of frictional losses due to air and rolling resistance, velocity and slope, together leading to continuous control of workload calculated in a power balance
(*b*) realistic simulation of the linear inertia of the wheelchair user system
(*c*) measurement of torques and forces exerted on rims as well as on the seat and backrest, during wheelchair propulsion
(*d*) possibility of analysing the effect of different wheelchair configurations
(*e*) possibility of measuring physiological and kinesiological parameters (movement analysis, electromyography).

At present various types of wheelchair ergometer design are used. These designs can be classified as follows:

(i) measurements on a track (GORDON, 1952; PEIZER *et al.*, 1964; LEHMAN *et al.*, 1974; GLASER *et al.*, 1980; SANDERSON and SOMMER, 1985), which are however hard to standardise, whereupon power output is difficult to quantitfy. Because the subject is not fixed to one place physiology can only be studied with portable devices and/or telemetry

(ii) wheelchairs used on a treadmill (VOIGHT and BAHN, 1969; ENGEL and HILDEBRANDT, 1973; DILLMANN and NIETERT, 1980; LESSER, 1986; VAN DER WOUDE *et al.*, 1986; 1989*c*). This experimental set-up allows proper analysis of physiology, kinematics and electromyography. Wheelchair propulsion is simulated in a most realistic way (inertia; steering required), although air resistance is absent. However, the latter is only important at relatively high velocities. Power output can be determined empirically in a separate drag test (VAN DER WOUDE *et al.*, 1986; 1989*c*). Forces and torques applied to the propulsion mechanism and/or seat and backrest can not, however, be studied in different wheelchair configurations

(iii) wheelchairs used on rollers (BROUHA and KROBATH, 1967; STOBOY *et al.*, 1971; BRAUER, 1972; MOTLOCH and BREARLEY, 1983)

(iv) wheelchair assemblies connected to a bicycle ergometer (BRATTGAARD *et al.*, 1970; WICKS *et al.*, 1977; GLASER *et al.*, 1980; FORCHHEIMER and LUNDBERG, 1986). (iii) and (iv) have the advantages of a stationary set-up. However, there is again the difficulty in measuring torques and forces. As in the treadmill and track analysis, different wheelchair configurations and anthropometry-based variations of the wheelchair are hard to study. Moreover, wheelchair propulsion is simulated with a simple mechanical or electric brake. Simulation of inertia (for instance with rotating disks) generally cannot be adapted to individual characteristics. Determination of power output can be inaccurate. This is especially important in short duration maximum effort tests (20–30 s). Isometric or isokinetic studies generally cannot be conducted on these devices

(v) seats with separate, instrumented rims (BRUBAKER *et al.*, 1981; JARVIS and ROLFE, 1982; LESSER, 1986; BURKETT *et al.*, 1987). These set ups do meet our demands to a certain but again still limited extent.

None of these designs fully comply with the demands in our first list. A complete simulation of the frictional losses in wheelchair driving is not possible in any of them. Wheelchairs used on a treadmill give good validity, but do not easily allow for either the exact calculation of workload or the measurement of forces on the hand-rims. Instrumentation to meet these requirements would be very costly, if different wheelchair designs are to be compared. Wheelchairs on rollers or combined with a bicycle ergometer need proper compensation for the inertial properties of the wheelchair/user system. Some ergometers have a provision for this (JARVIS and ROLFE, 1982, FORCHHEIMER and LUNDBERG, 1986; BURKETT *et al.*, 1987). Again, the instrumentation of the chair is expensive.

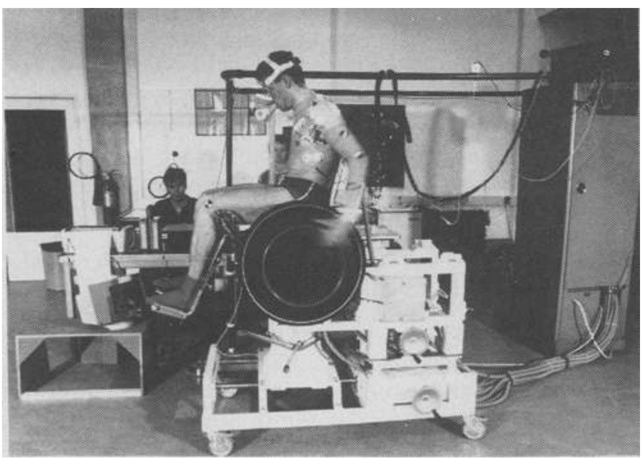

**Fig. 1** *The wheelchair ergometer in use in the biomechanical laboratory*

In a joint project with the Free University Amsterdam, the Erasmus University, Rotterdam has developed a stationary wheelchair ergometer (Fig. 1) which is designed to overcome the above problems. It comprises:

(*a*) a mechanical design which allows for the study of individual characteristics of the wheelchair/user interface. This design is highly adjustable for a wide range of different positions of handrims, seat and backrest positions and angles, without interfering with the instrumentation

(*b*) an electronic control system simulating frictional losses on the basis of feedback and also making possible isokinetic and isometric measurements. The rear wheels can be controlled separately, so that the simulation of wheelchair propulsion on a side slope is made possible

(*c*) a force measuring system allowing for the measurement of tangential, radial and medio-lateral forces of the hand on the handrims and for the measurement of reaction forces of the seat and backrest and the lower body and trunk

(*d*) this ergometer design will serve the detailed analysis of the wheelchair/user interface. Such a fundamental approach should ultimately lead to guidelines for wheelchair design and to a fitting model of the wheelchair to the user.

## 2 Controlling forces in hand-rim wheelchair propulsion

The most important of the controlling forces is the one the wheelchair user exerts on the hand-rim (Fig. 2). If $F_h$ is the resulting tangential force on the rim there will be a propelling force on the axis of the wheelchair

$$F_s = F_h \frac{R_h}{R_w} \tag{1}$$

in which $R_h$ and $R_w$ are the radii of the hand-rim and the wheel. As soon as $F_s$ overcomes the stationary rolling resistance the wheelchair will start to roll and from this moment on the following factors will have their influence on the acceleration and speed of the wheelchair:

(*a*) inertia of the wheelchair/user system
(*b*) rolling resistance
(*c*) air resistance
(*d*) slope.

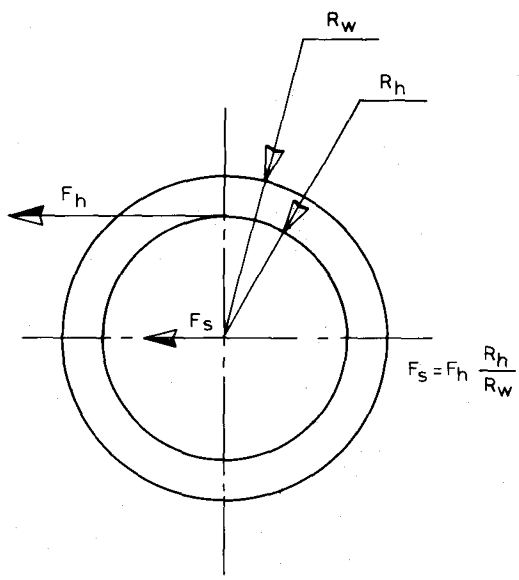


**Fig. 2** *Force on hand-rim results in propelling force $F_s$*

To express these factors in a formula of the form $F = ma$ we will use the following definitions:

(i) rolling resistance (Fig. 3) is a torque calculated from the total weight of the wheelchair user system $W$ and the imaginary distance between the weight vector and the normal reaction vector $N$. The force on the wheelchair as a result of rolling resistance is

$$F_r = \frac{W\varepsilon}{R_w} \tag{2}$$

In this formula $W$ and $R_w$ are constants for a wheelchair user system; $\varepsilon$ varies with tyre and road characteristics.

(ii) Air resistance is a force proportional to the square of the relative airspeed $(V_w - V_1)$ and the frontal area of the wheelchair user system $O$. It can be stated as

$$F_a = \tfrac{1}{2}\rho(V_w - V_1)^2 O C_d \tag{3}$$

in which
$\rho$ = density of the air (1·23 kg m$^{-3}$ at atmospheric conditions)
$V_w$ = wheelchair velocity m s$^{-1}$
$V_1$ = windspeed m s$^{-1}$
$C_d$ = resistance factor
The sign of $F_a$ is determined by the sign of $(V_w - V_1)$.

(iii) When the wheelchair is running on a slope there will be a force on the wheelchair

$$F_\alpha = W \sin\alpha \tag{4}$$

where $\alpha$ is the angle of the slope. An upward slope will be indicated as positive.

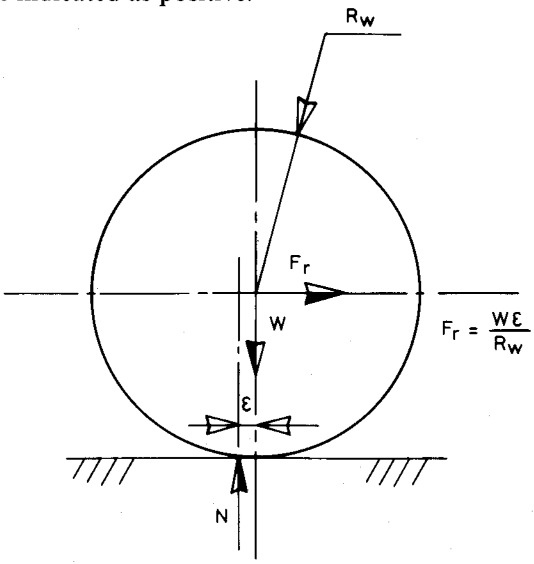


**Fig. 3** *Definition of rolling resistance $F_r$*

Now the resulting force $F_t$ on the wheelchair can be calculated with the equation

$$F_t = F_s - F_r - F_a - F_\alpha \tag{5}$$

and the equation of motion is

$$F_t = \frac{W}{g} a \tag{6}$$

where $g$ is gravitation and $a$ is the acceleration of the wheelchair.

This results in

$$a = \frac{F_t g}{W} \tag{7}$$

Now let us suppose we have a stationary wheelchair—a simulation of a real wheelchair—with free-turning wheels in which we continuously measure the torque $F_h$, also when the wheels are turning.

And let us suppose we have found values for $F_r$, $F_a$ and $F_\alpha$; then we can calculate at any moment the acceleration $a$ of the simulated wheelchair. Integration of $a$ according to

$$v = \int a \, dt \tag{8}$$

will give us the momentary speed of the chair.

To calculate $F_r$ we have to determine two variables, $W$ and $\varepsilon$. $W$ can be calculated from the weight of the user and an assumed weight for the wheelchair. During the test this is a fixed value. $\varepsilon$ can be varied during the test according to the road conditions to be simulated.

$F_a$ can be calculated with an estimated value for $O$, a given value for the windspeed, which may vary during the test, and a value for the drag $C_d$. This factor is approximately 1·4 (Coe, 1979). Further, we need a value for the wheelchair speed $V_w$, but this is exactly what we want to calculate. However, as these computations can be carried out by computer at a rate of approximately 50 per second, the latest calculated value for $V_w$ can be used here without causing a relevant error.

After a value for $\alpha$ has been set, $F_\alpha$ can easily be calculated. The slope angle $\alpha$ can also be varied during the test. Thus the acquired 'speed' of the simulated wheelchair is known at any moment. If we now have the wheels propelled proportionally to this calculated speed by means of speed-controlled servomotors we have created a realistic simulation of wheelchair propulsion under varying conditions.

## 3 Internal forces in the wheelchair/user system

Applying a force to the hand-rim will result in a reaction force on the user of the wheelchair. This force will be equalised by forces in the seat and the backrest of the chair. The study of these forces requires force transducers in the seat and backrest in frontal and vertical directions.

During actual propulsion the user exerts forces on the hand-rim which can be resolved into three directions: tangential, radial and axial. Only the tangential force will contribute to the propulsion of the wheelchair, the other two can be seen as losses. To study these forces three-directional force transducers will be required in the frame of the wheel suspension.

## 4 Measuring modes

Biomechanical research on human power output requires an analysis of three different ways of force generation

(*a*) realistic wheelchair simulation (isoinertial)
(*b*) isokinetic
(*c*) static.

Isoinertial measurements will be obtained by means of a realistic simulation of wheelchair propulsion. As described in the previous sections, the system reacts to the forces applied to the hand-rim. Isokinetic measurements are the study of manual forces at a constant speed of the wheels, which is not affected by forces exerted on the hand-rims. Static measurements in hand-rim wheelchair propulsion means the study of all forces while the wheels are totally blocked.

These three measuring modes complete the list of demands with which the ergometer must comply.

## 5 Stationary wheelchair design

### 5.1 *Mechanics*

Fig. 4 gives an impression of the design of the stationary wheelchair ergometer. Seat and backrest are mounted on a console (11) with a hydraulic foot-operated height control. The console itself can be moved in a forward or backward direction. In this way the chair can be positioned accurately with regard to the independently mounted wheels. The seat (9) and backrest (4) consist of two stiff frames, connected by three two-directional force transducers (10, 5) for frontal and vertical force measurement. Cushions of industrial manufactured wheelchairs can be mounted on the seat and backrest. Both seat and backrest can be tilted independently over 45°. Arm rests and legrests are adjustable and optional in use.

Next to the chair there are two side frames (1) for wheel suspension. Each side frame is mounted on three three-directional force transducers (12) (Fig. 5) for measuring all forces exerted on the hand-rims during propulsion. The wheels (3) (together with the driving motors) are mounted on subframes (2) to make possible a camber of the wheels of up to 15°.

The main suspension frames (1) can be moved sideways to adjust the width of the chair. Rims with different diameters can be mounted on the wheels. Between the hand-rims and the wheelchair-axis torque transducers (8) (GOMMERS, 1976; PRONK and NIESING, 1983) are mounted to measure the real torque that causes the propulsion. The motors (7) are 1000 W printed circuit motors (Mavilor 1000) mounted behind the wheel axis. Transmission is realised by timing-belts (6).

Tables 1 and 2 show the possible adjustments and the location and range of the force transducers.

### 5.2 *Electronics*

The electronic control unit (ECU) calculates the 'acceleration' and 'speed' of the wheelchair. Parameters for weight, air resistance, rolling resistance, acquired velocity and slope are taken from the software system (Fig. 6). The torque transducers give real values through a specially

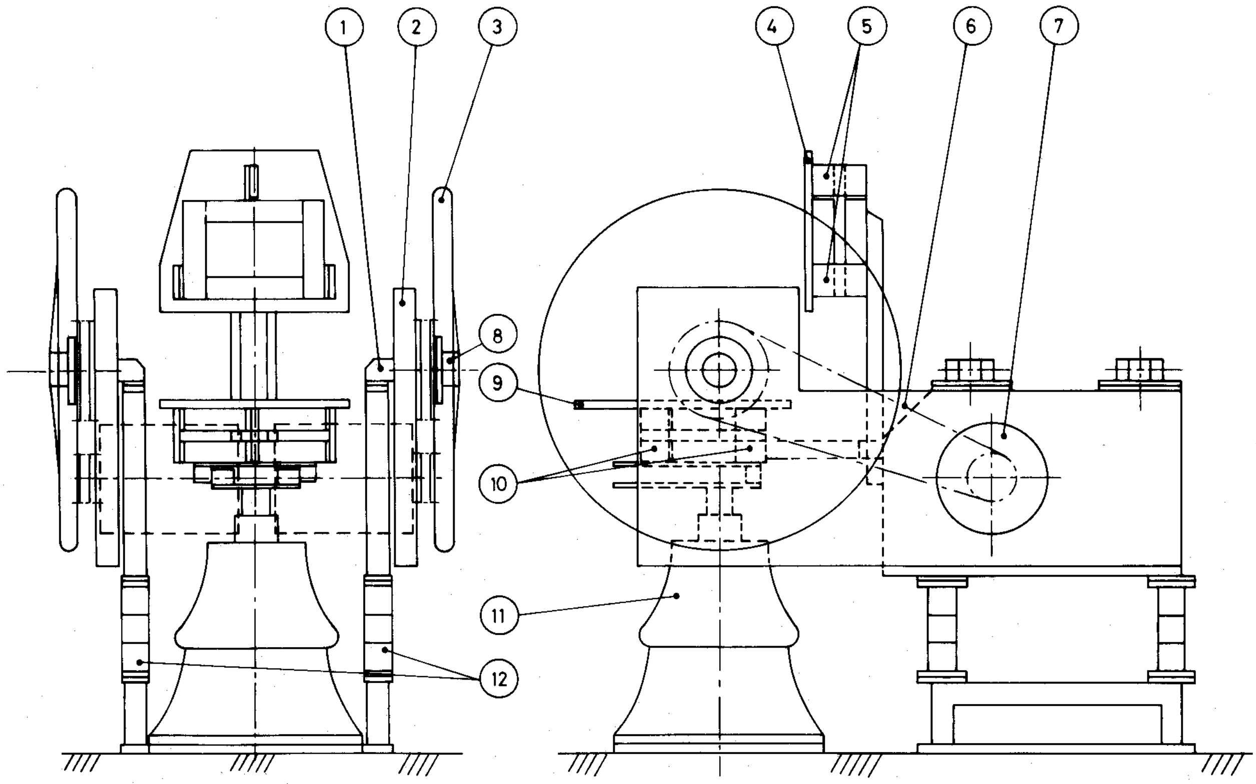


**Fig. 4** *Schematic front (left) and side view (right) of the ergometer. 1: side frames; 2: subframes; 3: wheels; 4: backrest; 5: two-directional force transducers; 6: timing belts; 7: motors; 8: torque transducers; 9: seat; 10: two-directional force transducers; 11: console; 12: three-directional force transducers*

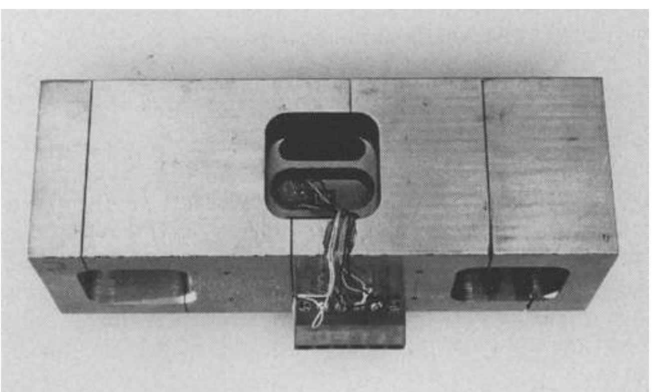

**Fig. 5** *Three-directional force transducer*

*Table 1 Adjustments of wheelchair dimensions*

| Adjustment | Range | Reference |
|---|---|---|
| Height control seat and backrest | −6–24 cm | wheel axis |
| Fore/aft alignment and seat and backrest | −22·5–10 cm | wheel axis |
| Tilting of seat and backrest | 0–45° | horizontal |
| Wheelbase (side frames) | 56–80 cm | |
| Camber | 0–15° | vertical |

*Table 2 Force transducers*

| Location | Number | Measured force | Range |
|---|---|---|---|
| Seat | 3 | perpendicular on seat (z) | 3000 N |
| Seat | 3 | in plane of the seat (x) | 750 N |
| Backrest | 3 | perpendicular on backrest (z) | 1150 N |
| Backrest | 3 | in plane of the backrest (x) | 1150 N |
| Wheels | 2 | torque handrim–wheel | 100 N m |
| Side frames | 2 × 3 | vertical force on side frame (z) | 850 N |
| Side frames | 2 × 3 | horizontal force on side frame (x) | 850 N |
| Side frames | 2 × 3 | horizontal force on side frame (y) | 850 N |

designed floating rotating measuring system. With eqns. 5–7 the actual 'speed' and thus the signals for the control amplifiers of the motors can be computed.

A second part of the electronic system contains the amplifiers for the 30 strain-gauge bridges for force measurement. Together with the torque transducers and the tachogenerators there are 34 variables which are sampled at 50 Hz. To handle this amount of information (20 400 bits $s^{-1}$) a data-acquisition unit (DAU) has been developed with a parallel connection to the PC-bus through a dual port RAM-memory.

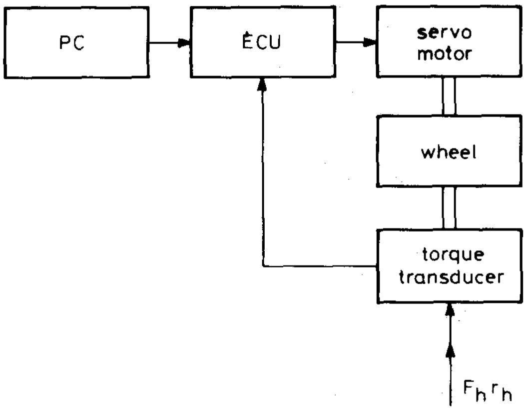


**Fig. 6** *Diagram of wheel speed control*

Stepwise multiple regression on all side frame transducers showed that only transducers measuring in corresponding directions contributed significantly to the applied calibration forces. The standard error of estimate stayed below 4·5 N for a gauging range of 400 N in all three directions. Subsequent tests showed independence of the position of forces applied to the rims. Moreover, the accuracy of simple linear regression equations using summated values of transducers in the *x*, *y* and *z* directions was shown to be comparable to the more complex multilple regression methods. Crosstalk of other transducers stayed below 3 per cent in all cases.

### 5.3 *Software*

The custom-made made software for use with the wheelchair ergometer contains four modules:

(i) administration
(ii) data communication
(iii) curve processor
(iv) processing utilities.

The administration module is a database containing files on the different test subjects/patients. The data used to define the experimental conditions (parameters for the frictional losses) are entered in the file. The data communication module gives real-time feedback for the subject on a screen containing information on the actual and required speed and direction of travel of the wheelchair. This module handles communications between the PC and the ECU (downloading of parameters), and between the DAU and the PC.

The curve processor is a program with the help of which the measured signals can be analysed interactively. A curve can be read in and processed. Processed data will be stored under a unique name, so that it is possible to repeat analysing steps. In the utility package curve processor commands are called from the command line. Series of calls can be collected in a batch file, enabling the user to access these command series by typing the single batch file name.

## 6 Preliminary results and discussions

The instrumentation in the analysis of a complex dynamic process such as manual wheelchair propulsion is of the utmost importance. In the study of the wheelchair/user interaction and the subsequent development of guidelines for propulsion technique and the geometry of the wheelchair/user interface, a combined physiological and biomechanical analysis is required. This required a testing device, which simulated wheelchair propulsion and allowed for the study of similar but simplified motions (isokinetics) and force application (isometrics). This led to the design and building of the stationary wheelchair ergometer presented in the previous part. The application of section such an experimental device lies within several areas of scientific interest:

(*a*) analysis of propulsion technique under both sports-like and daily-use conditions of wheelchair ambulation. Effective torque and mediolateral and radial hand forces give an indication of effectiveness of force application
(*b*) development of an (inverse dynamics based) segment model of the arm-shoulder-trunk. Forces generated on the propulsion mechanism in conjunction with kinematics and electromyography serve the calculation of net torques over the joints and the study of muscle co-ordination. Thus technique may be defined in more detail and it will serve the understanding of the 'human engine'. Subsequently, (over-)loading of joints or muscle groups can be studied. Together with physiological techniques, the variations in cardiorespiratory parameters between different wheelchair configurations can be studied in biomechanical detail
(*c*) the role of the seat and backrest in wheelchair propulsion and on the stability of the trunk in lower body disabled can be studied in detail.

Pilot studies were conducted to attain a first impression of the overall functioning of the prototype wheelchair ergometer, both under 'normal' daily-use conditions (VAN DER WOUDE *et al.*, 1989*a*; *b*) and under the more extreme conditions of a short-duration wheelchair sprint test (VEEGER *et al.*, 1989; VAN DER WOUDE *et al.*, 1989*d*).

In the first pilot experiment six male able-bodied subjects conducted three incremental exercise tests. The required velocity increased 0·28 m $s^{-1}$ every three minutes (0·56–1·66 m $s^{-1}$). Each test was conducted at different seat heights defined by elbow angle when sitting in an upright position with the hands on the top-dead centre of the rims (VAN DER WOUDE *et al.*, 1989*b*). The wheelchair/user interface, dominated by the position of the propulsion mechanism with respect to the seat/backrest, was adapted to individual anthrometric dimensions. These results were in line with previously results from a study of sitting height during wheelchair propulsion on a motor-driven treadmill and indicated an optimum seat height of around 100–200° elbow angle (VAN DER WOUDE *et al.*, 1989*b*).

In the second pilot study a number of wheelchair sprint tests were performed on the wheelchair ergometer at different friction loads (VEEGER *et al.*, 1989; VAN DER WOUDE *et al.*, 1989*d*). Different propulsion technique parameters were determined from the raw torque and velocity data. An individual curve of the power output (i.e. the product of momentary torque and angular velocity) at an intermediate level of friction is shown in Fig. 7. The sample frequency of 100 Hz used in this study allowed for within-cycle analysis of data. After recursive digital filtering using one of the curve processor options ($f_c$ = 20 Hz) each individual push (hand-to-rim contact) or recovery phase (period in which the hands progress from the end of

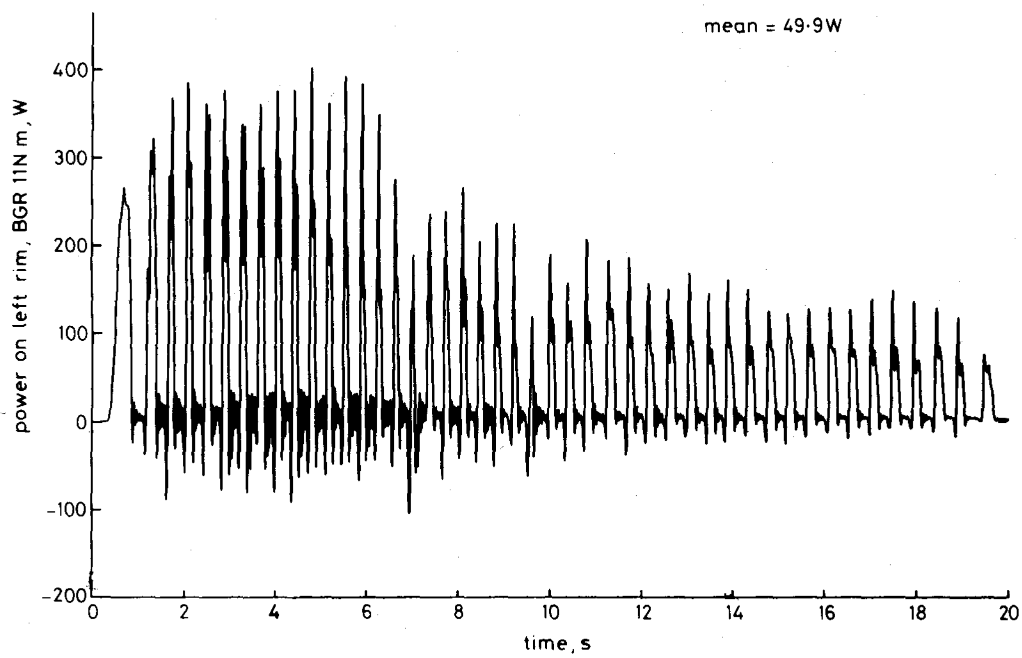


**Fig. 7** *Typical example of external power applied to the left hand-rim calculated following P(watts) = moment × velocity/rim radius*

the push phase to the initiation of the subsequent push phase) could be analysed. From the curve in Fig. 8 different timing parameters (duration of push, recovery and the complete cycle, time-to-peak torque) and characteristics of the power curve (peak power, peak of initial negative deflection due to imperfections in propulsion technique, mean values over the push and cycle period) can be derived.

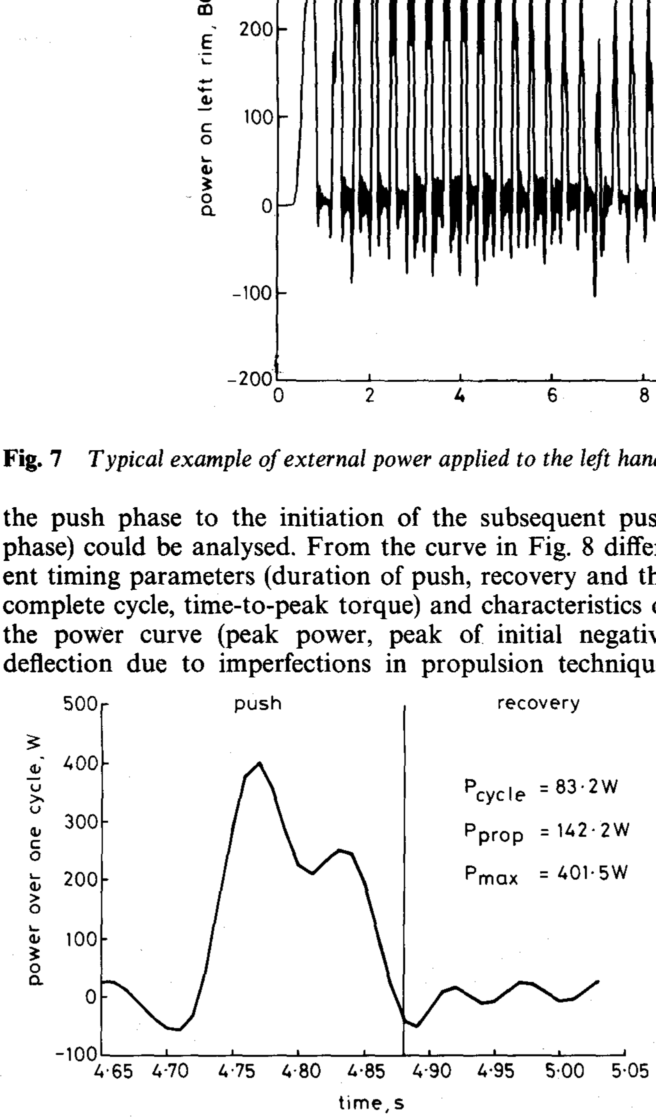


**Fig. 8** *Power output curve from Fig. 7 plotted over the propulsion cycle with the highest peak value*

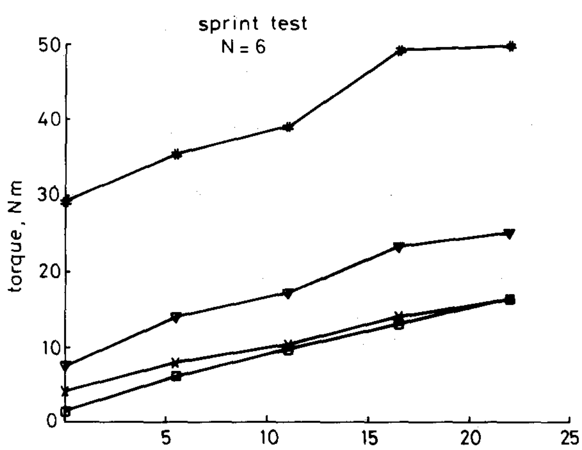


**Fig. 9** *Different torque parameters averaged over 15 s ($M_{15}$), over one complete propulsion cycle ($M_{cycl}$), over the push phase only ($M_{push}$), together with the measured peak value ($M_{peak}$)*
□ $M_{15}$ × $M_{cycl}$ ∇ $M_{push}$ # $M_{peak}$

Subsequently, group means may be determined for the different parameters and the dependency on friction load tested. Fig. 9 shows the effect of friction load on different torque parameters.

Preliminary results of a recent experiment actually comparing a group of subjects on both a motor-driven treadmill and the final design ergometer under theoretically equal conditions showed good agreement between cardiorespiratory data on both devices, indicating the validity of measurements using the wheelchair ergometer (Fig. 10).

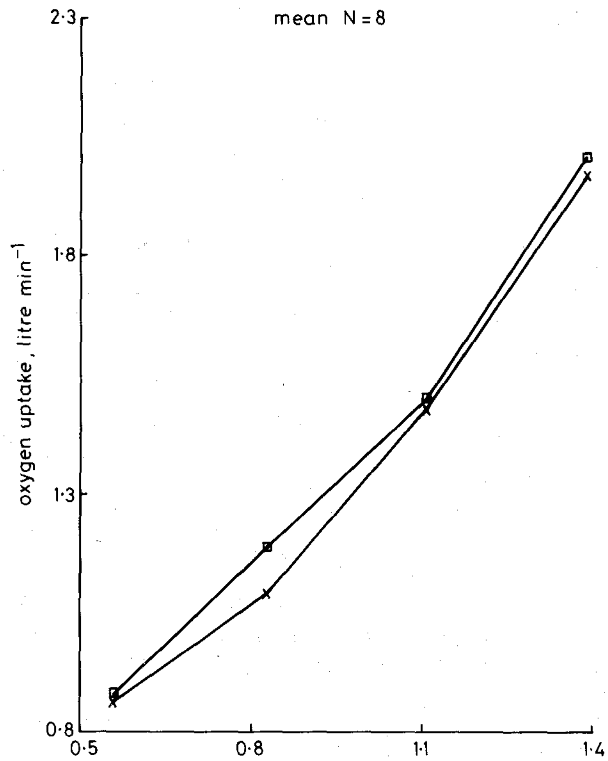


**Fig. 10** *Mean oxygen uptake values (N = 8) for propulsion on a motorised treadmill and under theoretically equal propelling conditions on the ergometer*
□ *treadmill* × *ergometer*

## 7 Conclusion

The presented computer-controlled stationary wheelchair ergometer is currently being used in experiments, studying both physiology and biomechanics of wheelchair propulsion. Such a fundamental approach is required to enable the development of practical anthropometry-based guidelines for propulsion technique and the design and fitting of the wheelchair to the user. Preliminary results indicated the functionality of the overall design. The computer-controlled system indeed allows realistic forms of wheelchair propulsion and the study of isokinetic and isometric conditions.

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## Appendix

*Detailed technical description of some parts of the ergometer*

1 *Force-measuring platform*

The force-measuring platforms in the seat and the backrest are constructed with three two-directional force transducers, making possible force measurements in the vertical and frontal directions. The strain-gauge transducer for one direction measures the shear force that results from a load on the platform, and is only sensitive for that one direction. The influence of forces and torques in other directions is low (GOMMERS, 1976; PRONK and NIESING, 1981). This property makes it possible to connect three transducers with stiff beams and still have little (interference) problem.

In this way a measuring platform is constructed with the following properties:

(i) stiff construction
(ii) electronic calibration before each measurement
(iii) measuring range 1–1000 N (resolution 1 N)
(iv) applicable in all positions from horizontal to vertical.

The three-directional transducers mounted under the wheel-suspension frames are based on the same principle.

2 *Electronic control unit*

The purpose of the electronic control unit (ECU) is to calculate and control the wheel speed from the given parameters (rolling resistance, air friction, slope) and the measured values (torque).

The ECU is assembled according to the block diagram (Fig. 11).

The A/D, microprocessor, memory and D/A blocks are assembled in a 19 in rack and communicate through the CRW bus. Both torque signals originate from floating transducers. The motor power unit is separately built in a 19 in rack and comprises a power supply, switching regulators and a feedback circuit. The tachogenerator, motor and gear are situated in the simulator itself. The centre of the ECU consists of a 16-bit microprocessor (Intel 8088). It takes care of the communication with the personal computer by means of a RS232 link and controls the A/D and D/A convertors. The microprocessor performs the necessary calculations, for which it has enough mathematical facilities (ADD, SUB, MUL, DIV).

To obtain the acquired calculating speed the firmware is written in assembly language. The firmware is stored in EPROM, together with test procedures. After the parameters have been downloaded from the PC to the microprocessor, a measurement session can begin. The firmware program analyses the parameters and determines the desired measurement mode (ISO-INERTIAL, ISOKINETIC, STATIC). A session is divided into a maximum of 15 blocks. Each block can contain different parameters for inertia of the wheelchair, rolling resistance, air resistance, required velocity and slope. The duration of each block is optional.

To minimise the amount of wireless electrical connections, the strain gauges of the torque transducers are measured with rotating strain gauge amplifiers. A voltage-to-frequency convertor modulates five infra-red light-emitting diodes which are mounted in the rotating part. Infra-red photosensors receive this signal in the stationary part. After these signals have been amplified, demodulation takes place in the torque measurement unit.

The electrical energy for power supply of the rotating part of the electronics is realised by means of a pair of rotating coupled air coils. The stator coil is powered with a sinusoidal audiofrequency signal. The rotator coil is tuned to this frequency. This gives the advantage of high efficiency and a low interference level.

### 4 *Data-acquisition system*

The data-acquisition system performs the following tasks:

(*a*) conversion of analogue voltages from the force transducers into digital values
(*b*) communication with the ECU by means of an RS232 connection
(*c*) communication with the PC by means of dual-ported RAM.

A single-board microcomputer forms the heart of the data-acquisition system. The A/D convertors are located on separate boards. These convertors are controlled by the microprocessor. The sample frequency is determined by a programmable timer. The timing process works on an interrupt base. The dual-ported RAM is located on an expansion board in the PC, and can be used by both the PC and the single-board computer, which communicates through an RS422 link.

To avoid interference signals, optocouplers are used. When one of the computers has written information into the dual ported RAM, the other computer is notified by means of an interrupt.

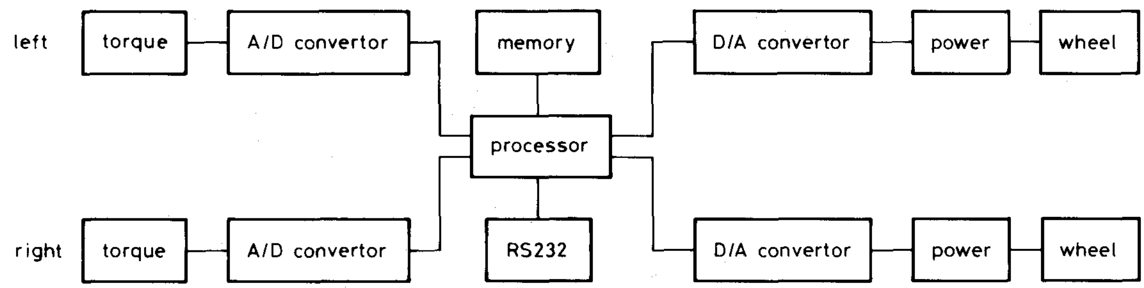



**Fig. 11** *Block diagram of ECU*

### 2.1 *Iso-inertial measurement*

In this mode the wheel speed is calculated from force, mass and the time during which the force is applied. For mathematical background refer to Section 2.

The firmware calculates the wheelspeed as follows:

(*a*) A/D conversion of torque signal
(*b*) conversion of torque of $F_s$ (eqn. 1)
(*c*) calculation and processing of dynamic losses
(*d*) determination of force direction
(*e*) processing of inertia
(*f*) integration
(*g*) guarding of limits
(*h*) conversion of the mathematical outcome into an analogue signal.

The above steps are cycled for both the right and left wheel once every 14 ms. The analogue signals are then fed to the motor control units. This process is illustrated in a flowchart (Fig. 12).

### 2.2 *Static and isokinetic measurement*

In the static and isokinetic mode fixed values for the wheel speed are presented to the D/A convertors. The motor control units converts these values into signals for each motor. The control loop formed by the tachogenerator, motor, gear and belt ensures a constant wheel speed. In the static mode this speed is zero.

### 3 *Floating rotating torque measurement*

To minimise miscalculation due to inertia of the moving parts the torque transducers must be mounted close to the hand-rims. Therefore, both torque transducers rotate with the hand-rims.

The demands can be summarised as follows:

(i) rotating measurement of positive and negative torques
(ii) no slide contacts, to ensure good reliability
(iii) separated system for left and right wheel
(iv) no electrical interference on ECG/EMG measurements.

### 5 *Safety*

Part of the firmware of the ECU is dedicated to safety. It controls and guards:

(*a*) maximum torque (both left and right)
(*b*) difference between required and actual speed
(*c*) software emergency stop.

Furthermore a manual emergency stop is implemented in the setup. High standards are demanded for electrical safety. These are met by the use of safety transformers, opto-couplers and shielded cables. The connections between the wheelchair and console are made by means of reinforced cables.

### 6 *Software specifications*

Four software modules support operation and experimental use of the wheelchair ergometer.

(i) administration module
(ii) data-communication module
(iii) curve processor
(iv) processing utilities

Modules (iii) and (iv) are applicable to other data-processing tasks as well.

It is assumed that the user of the programs is familiar with IBM-compatible PCs and the use of MS-DOS. The programs run on an Olivetti M24 or M28 (this because of the use of tiny text in graphic mode). The communication module uses a Labtender card of an ADSP (a CRW A/D convertor).

6.1 *Administration module*

The administrative data are stored in two databases. One contains data concerning the subject involved, the other contains data of the characteristics of a particular experiment. The programming environment is dBASE III$^+$. Assist (a menu-driven dBASE III$^+$ utility) enables the user to produce a variety of reports without the assistance of the programmer.

6.2 *Data-communication module*

The functions of the data-communication module are:

(i) feedback of speed-related information to the subject in the ergometer (real time). This is done by means of a 'steering wheel'. The height of the wheel represents the actual velocity and is presented together with the required speed. The angle of the 'wheel' with respect to the horizontal is a measure of

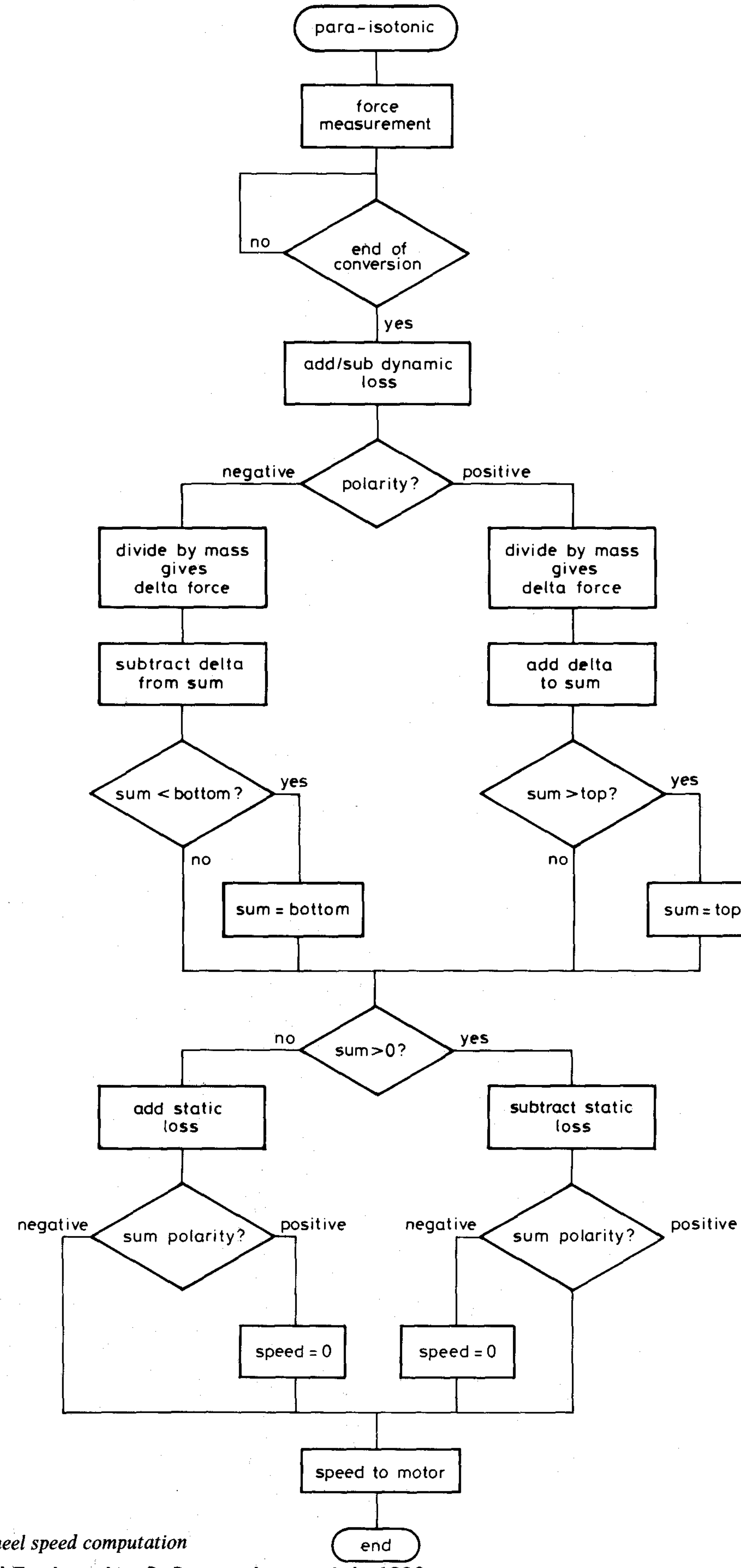



**Fig. 12** *Flowchart of wheel speed computation*

the speed difference between the left and the right wheel of the ergometer

(ii) measurement of 34 different signals:

(a) speed of both wheels (2)
(b) torque on both wheels (2)
(c) forces on back and seat of the chair (12)
(d) forces exerted on the wheel suspension frame (18).

(iii) real-time monitor of the 34 signals mentioned above (scope function)
(iv) storage of the measured signals
(v) the parameters to specify the experimental conditions are transferred to the electrical control unit (ECU) by the communication module.

The module has been written in Lattice C with the help of the Greenleaf Library.

6.3 *Curve processor*

The essence of this module is offline data processing. The module manipulates the data acquired from the communication module. A file is read in, processed and stored on disk again under a different name. In this way it is possible to process files in an iterative fashion.

The curve processor works interactively; its purpose is to test a certain idea. Bulk data processing is best performed with the utilities described in Section 6.4.

The program is 'open ended'. It enables the user to program their own routines in Turbo Pascal or GW Basic without leaving the curve processor. This makes it possible to create a personal library of data-processing modules. A large part of the programming necessary to communicate with the curve processor has already been done (in Pascal) and is available in the form of an include file. The following routines are implemented in the curve processor:

(a) detection of minimum and maximum values
(b) integration
(c) differentiation
(d) low-pass filter (Butterworth)
(e) linear fit
(f) fast Fourier transformation
(g) turbo Pascal link
(h) GW-Basic link
(i) MS-DOS link
(j) file type conversion (IEEE-real to ASCII and ASCII to IEEE-real).

The curve processor has been written in Lattice C with the help of the Greenleaf Library.

6.4 *Processing utilities*

All procedures are implemented within the curve processor (and some are also available in '.exe' form). These commands accept arguments on the command line, enabling the user to generate batch files. Batch files can be processed at those times when the computer is generally not in use (e.g. at night), without human assistance. They are also suited to perform complex tasks that are to be repeated several times. The commands that make up the task are stored in a batch file. Only one command then activates this function.

Knowledge of the MS DOS batch file processor is useful (e.g. to implement loops and conditional jumps).

6.5 *Portability of measurement data and results*

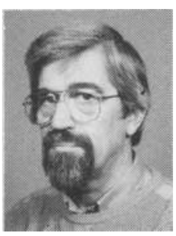
The data format used in all files is the IEEE-real format. The data in each file are uniform; there is no header information. This makes possible an easy conversion to ASCII, permitting transport to other programs.

## Authors' biographies

Ruud Niesing was born in 1941 in Rotterdam, The Netherlands. He graduated from the Faculty of Mechanical Engineering of Delft University of Technology in 1970. He joined the Central Research Laboratories of Erasmus University Rotterdam in 1971 as a designer and is now head of the Mechanics & Glass Department. His main interests are force-measuring equipment and electromechanical devices.

Frits Eijskoot was born in 1947 in Rotterdam, The Netherlands. He joined the Central Research Laboratories of Erasmus University Rotterdam in 1968. He is working as an electronic designer, and his major interests are in the field of biomedical electronics in general and sensor application in particular.

Ries Kranse was born in 1960. He joined the Central Research Laboratories of Erasmus University Rotterdam as a computer programmer in 1985. At present he is involved in research in the Department of Urodynamics.



Arie den Ouden was born in 1936 in Rotterdam. He acquired much experience in several jobs in industry before he joined the Central Research Laboratories of Erasmus University Rotterdam in 1969 as a designer. He is mainly involved in physiological measuring equipment and aids for the disabled.

Joop Storm was born in 1946 in Dordrecht, The Netherlands. In 1967 he was one of the first to join the newly established Medical Faculty at Erasmus University Rotterdam. He is currently working as a senior designer and project engineer at the electronics department of the Central Research Laboratories. His major interests are project engineering, mechatronics and horses.

Dirk-Jan Veeger has an M.Sc. in Human Movement Sciences as well as Ergonomics. He is research fellow in the Department of Functional Anatomy of the Faculty of Human Movement Sciences of the Free University Amsterdam, The Netherlands.

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Chris Snijders was born in 1941 in Malang, Indonesia. He graduated from Eindhoven University of Technology in 1966 and acquired his Ph.D. in 1970. After being a member of the scientific staff of the Department of Mechanical Engineering at Eindhoven University of Technology (1966–1984) and Twente University of Technology (1976–1984) he became a Professor in Medical Technology at the Faculty of Medicine of Erasmus University Rotterdam in 1984 and at the same time at the Department of Measuring & Control & Ergonomics of the Faculty of Mechanical Engineering of Delft University of Technology.