TECHNICAL NOTE

A NEW UROFLOWMETER—AN INSTRUMENT FOR THE ESTIMATION OF LIQUID VOLUME AND RATE OF FLOW

INTRODUCTION

The determination of urinary flow parameters is essential in certain studies of the urinary system. The apparatus described was designed to assist in assessing the effect of treatment for relieving obstruction of the lower urinary tract. The requirement was for a portable instrument able to simultaneously record both rate of flow and total volume, whilst allowing privacy to the patient during voiding. In addition the design had to ensure simplicity of operation and ease of cleaning.

There are numerous reports in the literature of instruments having a similar purpose. Thus Drake (1948) has described a simple uroflowmeter improved later by Kaufman (1957), which detected the weight of urine voided by a spring balance arrangement and recorded total volume voided on a kymograph. Von Garrelts (1956) detected the volume of urine collected in a tall cylinder by means of a pressure transducer at the base. Both volume and rate of flow were recorded; the latter being derived electronically. The components were relatively expensive and the essential electronics complex. Holm (1962) used the flow of urine to displace air and measured the air flow rate using an anaesthetic gas flowmeter. There was no means of recording. Cardus et al. (1963) used an electromagnetic blood flowmeter to measure the rate of flow of urine through a funnel connected to the instrument's probe. Koontz and Rowan (1967) employed a carousel of 50 ml containers, which appeared in succession below the collection funnel at a rate of one a second. The rate of flow and volume voided were calculated later from the measured contents of the separate containers. None of the previous designs individually fulfil all the above mentioned requirements.

APPARATUS

The complete apparatus shown in Fig. 1 is made up from three separable units. The urine collection receptacle is connected to electronic and recorder units by 15 ft (4.6 m) of flexible cable. The equipment can thus be arranged to afford the patient complete privacy, minimizing psychic inhibition, whilst the operator retains overall control of the recording. Urine is passed via a wide necked funnel into a standard 500 ml glass beaker, which rests in a removable pan. The latter is supported by a stiff cantilever, which forms part of a commercially available strain transducer.* The load in the pan is transmitted to the cantilever by a vertical steel spindle in a P.T.F.E. sleeve (Fig. 2). Thus the weight of the accumulated volume of urine is converted into a proportional strain in the cantilever, and is detected by a pair of attached semiconductor strain gauges, which form two arms of a Wheatstone bridge arrangement. The resultant electrical output is fed to the electronic unit, which provides two separate outputs to the twin channel chart recorder. The latter is of the simple moving coil type. Deflection in one channel is proportional to accumulated volume with a full scale deflection of 500 ml, whilst deflection in the other channel is proportional to rate of flow with two switched ranges having full scale deflection of 50 and 25 ml s⁻¹. The chart speed is 8 in. (203 mm) min⁻¹, and Fig. 3 shows a typical tracing.

Details of the electronic circuit appear in Fig. 4. Resistors R2 and R5 form the other two arms of the Wheatstone bridge. Resistors R1, R3 and R4 together with the 5 turn potentiometer VR1 provide a means of rendering zero output from the bridge, when the beaker is empty. This SET ZERO control is offset and of wide range to accommodate differences in weights of beakers. The resultant bridge output current is amplified by the integrated circuit amplifier A1. The gain determined by the ratio of R13 to R5 is set so as to provide approximately 4.5 V at the amplifier output for a load of 500 ml. A nominal 1 mA current is supplied to the volume channel of the recorder through resistor R12 and sensitivity adjustment potentiometer VR2. Capacitor C5 in parallel with R13 causes the response of the amplifier to fall off above 90 Hz, thereby reducing the effect of shock and vibration at the detector.

A voltage proportional to rate of flow is obtained by electronic differentiation of the output of amplifier A1. Amplifier A2, connected in the inverter configuration, provides a very low input impedance to the charging current of C6, which is proportional to the rate of change of volume. The rate recorder channel is supplied from either of two switched resistor chains connected to A2 output, giving full scale sensitivities of 25 and 50 ml s⁻¹. Components C6 and R19 are chosen to give the desired sensitivity. C13 and R14, are included to attenuate the upper frequency response of the differentiator, which would otherwise have an undue sensitivity to circuit noise. The turnover point in the response was chosen to be equivalent to a time constant of 100 ms. From experiments this was deduced to be the best compromise in that significant changes of rate were still recorded, whilst sensitivity to turbulence was kept low.

* Available from Devices Instruments Ltd.
The resistance bridge energization potential of 5.6 V is stabilized by zener diode Z3. The amplifier power supplies of plus and minus 15 V are provided by the emitter followers T1 and T2, which use zener diodes Z1 and Z2 as reference.

The urine collector is mounted on a vertical stand and can be adjusted to the most convenient height for the patient. A push button switch mounted above the collector is pressed by the patient just before start of voiding. This action initiates movement of the recorder charts and prevents wastage of chart paper. The recorder is stopped by turning the control switch on the electronic unit from READY to OFF. In the ON position of this switch the recorder chart progresses independently of any action of the patient.

CALIBRATION

The volume channel is calibrated by observing the recorder deflection produced by adding carefully measured volumes of water to the collector. Non-linearity in the system is almost wholly due to the recorder, but is in any case less than 1 per cent of full scale deflection. Calibration of the rate channel is made relative to the volume channel. Thus necessary adjustments in sensitivity, to take account of different fluid densities, need be made to only the volume channel. As constant rates of flow of water are difficult to arrange, it was found more convenient to simulate a change in volume by inserting a linearly increasing current between the output terminals of the resistance bridge. The current was derived from a slow sweep ramp voltage generator connected through a