STUDIES WITH A NEW PUMPLESS, PARALLEL-FLOW HAEMODIALYZER*

A. R. LAVENDER**, FINLEY W. MARKLEY and MARVIN FORLAND

Department of Medicine, University of Chicago, Chicago; Argonne Cancer Research Hospital (operated by the University of Chicago for the U.S. Atomic Energy Commission) and Argonne National Laboratory (operated under the auspices of the U.S. Atomic Energy Commission), Argonne, Ill., U.S.A.

Desirable features of a haemodialyzer suitable for home use are simplicity of operation, small size, low priming volume, and low cost. The simplest dialyzer is one which requires no blood pump. Size and priming volume may be reduced by using thin blood films and thin membrane supports. Cost may be minimized by a design adaptable to high speed, automated manufacturing techniques.

A pumpless unit must have low flow resistance, a requirement which necessitates a parallel-flow configuration. The problem then becomes one of manifolding multiple channels in parallel.

With the exception of the hollow fibre dialyzer (Lipps et al., 1967), manifolding usually is accomplished by means of inserts which are placed between two sheets of cellulose membrane. A leak-proof seal is provided by clamps or bolts which compress membrane against inserts. All parts must fit accurately in order to prevent leaks.

The numerous inserts in multilayered dialyzers may account for a significant fraction of flow resistance and may increase dialyzer dimensions. The thickness of the membrane support is the primary determinant of a dialyzer's dimensions, and support thickness depends in part upon thickness of manifold inserts. In order to preserve dimensional stability and uniform flow distribution, support thickness must match insert thickness. It follows, therefore, that elimination of manifold inserts will reduce the size of the dialyzer.

We have manifolded a parallel-flow, multilayered haemodialyzer by a technique which does not require tubing inserts or compression seals (Lavender et al., 1968). The unit was constructed from standard dialysis tubing, inexpensive membrane supports, and a plastic housing. Design details and results of preliminary tests are presented here.

MATERIALS AND METHODS

**Membrane support.** The support was a plastic screen mesh of the type described by Hoeltzenbein (1966). Mesh filaments of 0.5 mm diameter were arranged in a diamond patterned lattice work. The screen thickness was 1.0 mm. Each layer of flat cellulose tubing was interposed between two layers of mesh. Blood channel height was determined by filament diameter and by membrane sag between support points on the mesh. Uniformity of blood channel height was a function both of filament diameter uniformity and of mesh pattern uniformity. Approximately 67 per cent of the membrane area was exposed to blood and dialysate.

**Dialyzer (Fig. 1).** Eighteen cm lengths of 4.5 cm wide, flat, cellulose dialysis tubing were stacked alternately with strips of membrane support. The support strips were slightly shorter

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and wider than the tubing. A forty layer stack provided 5400 cm² of membrane area, 3600 cm² of which were exposed to blood and dialysate. Both cellophane and cuprophane tubing were used.

The tubing ends were opened, fixed at a channel height of 0.125 mm, and imbedded in epoxy resin. The resin filled the interstices of the membrane supports for a distance of 1.5 cm and separated the dialysate compartment from the blood compartments. The cut ends of the tubing were buried in the resin and transition from resin to cellulose was smooth. When the dialyzer was wetted, swelling of the cut ends locked them inside the resin. The resin maintained the blood channel height of the tubing entries, and was molded to form a curved entry into each tube.

The stacked, potted unit was placed in a methylacrylate housing which was fitted with entry and exit tubes for blood and dialysate. Shims were placed between the housing and outer membrane support on both sides of the dialyzer. The shims, by bringing all tubing and mesh layers into tight opposition, reduced blood volume and created multiple blood channels in the mesh interstices.

Silicone rubber was poured onto the edges of the stacked layers. The polymer flowed to the edges of the cellulose tubing layers and obliterated all space between the housing and the tubing. One short segment on each side of the dialyzer was kept free of rubber and served as an entry or exit for dialysate into the stack. Dialysate entered the dialyzer in the segment adjacent to the blood exit chamber. Dialysate left the dialyzer in the segment adjacent to the blood entry chamber. A tube from the dialysate compartment was used to monitor dialysate chamber pressure.

The overall dimensions of the completed dialyzer were 23 × 7 × 6.5 cm. The functioning part of the unit was 15 × 5 × 4.5 cm. The dry weight of the dialyzer was 770 grams and much of the weight was attributable to the heavy plastic housing.
Protocol. The dialyzer was in a vertical position in all experiments. Blood was propelled into the top by a constant speed roller pump. A single pass system was used and dialysate flow always was countercurrent to blood flow. Blood and dialysate flows were measured by timed collections of outflow in a graduated cylinder.

A screw clamp on the blood outflow line regulated inflow and outflow blood pressures. Dialysate chamber pressure always was subatmospheric and was regulated by a screw clamp on the dialysate inflow line. Dialysate temperature was maintained constant at 39°C.

For in vitro experiments, a modified Ringer's solution with added creatinine and urea served as blood. Tap water dialysate was used. For in vivo experiments, standard isomotic dialysate solution was used. In all studies, blood entering and leaving the dialyzer, and dialysate entering and leaving, were sampled simultaneously for analysis and estimation of dialysance.

In animal experiments, mongrel dogs of 18 to 20 kg body weight were anaesthetized with intravenous pentobarbital sodium. The femoral artery and vein were cannulated and were connected to the pump and dialyzer by silicone rubber tubing. All animals were heparinized. A constant speed pump infused creatinine and urea intravenously.

Ultrafiltration of aqueous solutions was determined by collecting filtrate from the dialysate compartment while pumping through the blood channels at a constant pressure.

Chemical methods. Creatinine was estimated by a modification of the method of Bonsnes and Taussky (1945). Urea was determined by the method of Coulombe and Favreau (1963). Sodium and potassium were estimated by flame photometry. Dialysance was calculated from:

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\text{Dialysance} = \frac{Q[(A - V)/A]}{Q}, \quad \text{where,}
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Q = \text{blood flow}
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A = \text{input blood concentration}
\]
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V = \text{output blood concentration}
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RESULTS

Characteristics of the dialyzer. Surface area of a 40 tube dialyzer was 5400 cm², of which 3600 cm² was exposed to blood and dialysate. Blood plenum chamber volume was 50 ml. Volume of the blood channels was 40 ml at zero transmembrane pressure. Volume doubled when transmembrane pressure was increased to 300 mm Hg. At zero pressure the blood channel height averaged 0.177 mm.

The membrane did not rupture or leak at resin-tubing seals when transmembrane pressures were as high as 600 mm Hg. The ultrafiltration rate of aqueous solutions was 180 ml/hr at a filtration pressure of 300 mm Hg and a flow rate of 300 ml/min.

Resistance to flow was low. Pressure drop in the dialyzer was 2 to 5 mm Hg when aqueous

![Graph](image-url)

**Fig. 2.** Effect of blood flow on dialysance in vitro. Cuprophan dialyzer; 5400 cm²; dialysate flow 520.
solutions were pumped at 300 ml/min. with an input pressure of 100 mm Hg. Pressure drop was 10 to 20 mm Hg at similar flows and pressures when dog blood with haematocrit values of 35 to 50 per cent was pumped through the dialyzer. When the dialyzer was connected by 6 mm I.D. tubing to the femoral artery of a dog with a mean arterial pressure of 100 mm Hg, flow through the dialyzer exceeded 1200 ml/min.

Aqueous solutions were pumped continuously through the blood and dialysate compartments for as long as one week without development of leaks. Dialyzers rinsed free of blood and stored in the refrigerator were re-used several times over intervals as long as one month.

**In vitro tests** (Fig. 2). When dialysate flow was held constant at 520 ml/min., dialysance increased as blood flow increased. At a blood flow of 290 ml/min., urea dialysance was 90 ml/min. and creatinine dialysance was 60 ml/min. Sodium dialysance paralleled urea dialysance and potassium dialysance was 110 ml/min.

When blood flow was held constant, dialysance increased as dialysate flow increased (Figs. 3 and 4). Dialysance approached a limit at a dialysate flow of 400 to 600 ml/min. When dialysate flow was 540 ml/min., creatinine and sodium dialysances were 45 ml/min. and 75 ml/min., respectively, in a cellophane dialyzer (Fig. 3). When cuprophan was used, sodium dialysance was 80 ml/min. and creatinine dialysance was 52 ml/min. (Fig. 4). Peak values for sodium and creatinine dialysances were 75 and 55 ml/min., respectively, for cellophane, and 90 and 68 ml/min., respectively, for cuprophan.

**Fig. 3.** Effect of dialysate flow on dialysance *in vitro*. Cellophane dialyzer; 5400 cm²; blood flow 372.

**Fig. 4.** Effect of dialysate flow on dialysance *in vitro*. Cuprophan dialyzer; 5400 cm²; blood flow 360. Compare with Figure 3.
The effects of storage on dialyzer function are shown in Figure 5. This cellophane dialyzer was tested on two occasions, one month apart. The unit was stored in the refrigerator between testing periods. Dialysances of sodium and creatinine were the same on both occasions.

*In vivo tests.* At a constant blood flow dialysance increased with dialysate flow, but approached a limit at 400 to 500 ml/min. dialysate flow (Table I).

Blood was pumped through dialyzers for 6 to 8 hours. A clamp on the outflow line maintained outflow pressure at 80 to 100 mm Hg. Clotting did not occur in the dialyzer and flow resistance did not increase during an experiment. Blood was rinsed from the dialyzer with tap water at the end of each experiment.

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<th>Blood flow ml/min.</th>
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DISCUSSION

The design objectives for the dialyzer were compactness, low priming volume, low flow resistance, and low dialysate consumption. These objectives have been achieved and we are hopeful that the unit can be fabricated inexpensively on automated machines.

Urea and creatinine dialysances of cellophane dialyzers were approximately one-third of values obtained by us with plastic screen twin-coils which had a surface area of 14,500 cm². When blood flow through these coils was 310 ml/min., urea dialysance was 184 ml/min. and creatinine dialysance was 100 ml/min. When cuprophane tubing was used in the parallel-flow dialyzer, performance improved to approximately one-half of that of the twin-coil.

Maximum dialysance was achieved at low dialysate flows, which suggests that membrane scrubbing by dialysate was efficient. The major resistance to solute transfer was probably on the blood side of the membrane. Blood was unevenly distributed across the membrane, as shown by dye dilution curves and rapid sequence photography. Dye injected into the blood inlet line traversed central channels more rapidly than those in the periphery, and channelled in the outer layers of the dialyzer. Dye appeared more rapidly and disappeared more slowly than was predicted by dialyzer length, volume and blood flow.

Predicted linear velocity was 0.95 to 1.90 cm/sec., whereas velocity in the twin-coil was approximately 28 cm/sec. This low velocity probably impaired mixing in blood channels. Indirect evidence for this was the difficulty with which air was purged from the dialyzer. Small bubbles which adhered to projections from the mesh were difficult to remove, in contrast to the ease with which they can be removed from twin-coils wrapped with plastic screen membrane support.

The dialyzer's geometry is now being optimized to improve performance. The blood plenum chamber is being reduced in size to decrease priming volume. The blood plenum chamber configuration is being altered to improve flow distribution. Surface area is being increased by use of thinner membrane supports. We have built, but not evaluated, an 80 tube unit with 10,800 cm² surface area. Membrane supports of 0.5 mm thickness were used. Modification of the support should permit further reduction in size and priming volume.

Although a number of problems remain to be solved, this dialyzer should be useful clinically in its present form. The cuprophane unit is about equal in performance to a two layer Kii1 and has about 50 per cent of the performance of a plastic screen twin-coil. The low priming volume and small change in volume with pressure should be advantageous for dialysis of infants and small children. The dialyzer's simplicity and ease of operation may be of value in treatment of acute hyperkalaemia and drug intoxications. The compactness of the unit and its low flow resistance permit mounting of the dialyzer close to the patient without a blood pump. Blood losses in tubing are thereby minimized.

Summary

A new pumpless, parallel-flow haemodialyzer is described. A dialyzer with a surface area of 5400 cm² has a priming volume of 90 ml. Flow resistance was minimal and blood clotting was not a problem. The performance data and design features suggest that the dialyzer will be useful for future applications in clinical haemodialysis.

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REFERENCES


