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SYNTHETIC HYDROGELS IN CONTACT LENS APPLICATIONS

by

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A thesis submitted for the degree of

Doctor of Philosophy

in the

University of Aston in Birmingham

January 1974

SUMMARY

The aim of the work described in this thesis is the correlation of the structures of some (homogeneous) synthetic hydrogels with those properties relevant to contact lens applications. The significance of the results in contact lens wear as judged from limited clinical trials is also discussed.

A hydrogel is a flexible, covalently cross-linked polymer network in a water swollen state, having a glass transition temperature lower than room temperature. Hydrogels based on homopolymers, copolymers and terpolymers were studied in this work. The effects of chemical structures and compositions of the components, and cross-link density, together with polymerisation conditions, hydration temperatures and sodium chloride concentration (of the hydrating solutions) on the equilibrium water contents of the hydrogels were investigated. Thus sheets of a variety of hydrogel materials based on copolymers of a hydrophilic and a hydrophobic component and of two hydrophilic components, as well as terpolymer sheets of two hydrophilic components and one hydrophobic component were prepared by bulk polymerisations. The hydrophilic monomers used were hydroxyethyl methacrylate, hydroxypropyl methacrylate, hydroxyethyl acrylate, hydroxypropyl acrylate and vinyl pyrrolidone, while the hydrophobic monomers used were styrene, para-methyl styrene and methyl methacrylate. The copolymer sheets were characterised by density and refractive index measurements in both dehydrated and hydrated states.

Techniques were developed to enable the effects of chemical structures and equilibrium water contents of the hydrogels on those physical properties believed to be most relevant to contact lens applications, such as oxygen permeability and deformation properties, to be studied. A high-water-content hydrogel was found to have a lower density, lower refractive index, higher "dissolved" oxygen permeability

coefficient and larger deformation under a load, than a low-water-content hydrogel. The chemical structures of the polymers also affected the oxygen permeability and deformation properties in the low-water-content hydrogels (with equilibrium water contents less than 40%): a hydrogel containing more flexible components had a higher "dissolved" oxygen permeability and gave a larger deformation under a load, than a hydrogel containing more rigid components.

"Dissolved" oxygen permeability coefficients of the hydrogels were found to increase exponentially with their equilibrium water contents over a wide range. Three possible states of water (i.e. "primary", "secondary" and "liquid-like") were suggested.

It was predicted that for a 0.1 mm thick continuous-wear contact lens, the hydrogel should have "dissolved" oxygen permeability coefficient of 230×10^{-10} cc(STP).mm.cm^{-2}.sec^{-1}.cm Hg^{-1} at 34°C, i.e. a minimum equilibrium water content of 52% in 0.9% saline at 34°C (based on the literature values of 15 mm Hg for the critical oxygen tension at the corneal surface and 3.5 µl.cm^{-2}.hr^{-1} for the minimum oxygen flux into the corneal epithelium, in order to maintain corneal transparency).

A preliminary investigation into preparing polymer rods by bulk copolymerisation of some of the monomers for subsequent lathe-cutting trials, and some polymer powders by solution polymerisation for subsequent compression moulding trials were also carried out.

This work was carried out between 1970 and 1973 in the University of Aston in Birmingham. It has been done independently and has not been submitted for any other degree.

C. O. Ng.

ACKNOWLEDGEMENTS

I wish to express my sincere thanks to Dr.B.J.Tighe for his constant guidance, concern, encouragement and hours of discussions throughout the course of this work.

I should also like to thank Dr.J.R.Larke of the Department of Ophthalmic Optics for some helpful discussions on the contact lens field, and his limited clinical results.

My thanks go to the other members of the Chemistry Section of the research team, Dr.R.E.Ensor and Mr.D.G.Pedley, for their help during the period of this work, and also to some members of the laboratory staff in the Department of Chemistry, Mr.M.H.Hughes in particular, for their technical help such as sample analysis.

I am very grateful to the National Research Development Corporation (i.e. N.R.D.C.) and Kelvin Lenses Limited for their financial support of this work.

Finally, I would like to thank Mrs.Stephenson for typing this thesis.

ABBREVIATIONS

HEMA	:	hydroxyethyl methacrylate
HPMA	:	hydroxypropyl methacrylate
GMA	:	glycerol methacrylate
MMA	:	methyl methacrylate
HPA	:	hydroxypropyl acrylate
HEA	:	hydroxyethyl acrylate
St	:	styrene
pMS	:	para-methyl styrene
4MPl	:	4-methyl pent-1-ene
EA	:	ethyl acrylate
VP	:	N-vinyl pyrrolidone
EGDMA	:	ethylene glycol dimethacrylate
BzP	:	benzoyl peroxide
AZBN	:	α-azo-isobutyronitrile
E.W.C.	:	equilibrium water content
W.C.	:	water content
"DO"	:	"dissolved"oxygen
"GO"	:	"gaseous" oxygen

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GENERAL INTRODUCTION

Like spectacles, a contact lens is an optical appliance for correcting refractive errors of the eye. But unlike spectacles, a contact lens is in contact with the cornea, as the name implies. Mainly for cosmetic reasons and for convenience more and more people, especially women and sportsmen, have become contact lens wearers. The population of contact lens wearers is small, however, although contact lenses have been accepted since 1948, and was estimated to be only about three percent of spectacle wearers in 1972. Among the many reasons which account for this are included adverse effects of present contact lenses on the cornea (such as comfort, visual acuity, edema, etc.) and the high cost of a pair of contact lenses (£25-60 for a pair of hard contact lenses, and £65-100 for soft contact lenses, and about two-thirds of which goes to the optician who fits the lenses). The high cost is mainly attributed to the time consumed by the optician in fitting and many following check-ups over a period of time.

Since 1938, contact lenses have been made of plastics or elastomers. By 1970 three types of contact lenses were known, namely, poly(methyl methacrylate) [i.e. PMMA], poly(2-hydroxyethyl methacrylate) hydrogel [i.e. PHEMA] and silicone rubber [i.e. poly(dimethyl siloxane)]. PMMA contact lenses (so called hard lenses) have been worn since early 1950, whereas PHEMA hydrogel contact lenses (so called soft lenses) and silicone rubber lenses were developed in the past decade and were undergoing extensive clinical tests. However these materials are not satisfactory in one way or another, and the contact lenses are accepted for daily wear only (as distinct from continuous wear). Daily wear contact lenses are subject to three main disadvantages from the wearer's point of view, namely, (a) contamination during daily handling on the lens,

(b) rupture or short life span due to mishandling of the lens and (c) higher chance of losing the lens. To reduce these shortcomings in daily wear contact lenses, the idea of contact lens for continuous wear has been suggested. In order to develop a new contact lens material, bearing in mind the requirements of contact lens wear in terms of comfort and vision, and particularly the physiological needs of the cornea, which will enable the lens to be worn in the eye continuously during successive day and night periods without being harmful to the eye, a research programme was initiated in October 1970 in the Department of Chemistry working in conjunction with the Department of Ophthalmic Optics and sponsored by the National Research Development Corporation and Kelvin Lenses Limited. The role of the Chemistry section in the whole research programme is to synthesise polymers and to correlate their structures with those properties relevant to contact lens applications in order that contact lens materials (preferably synthetic hydrogels) of high oxygen permeability to satisfy continuous wear requirements will be ultimately developed. In the past three years during which this research work was carried out, some new contact lenses appeared in the market, but this did not diversify to any appreciable extent the line of work in this research. This work was based on the literatures[1,2] available in October 1970 and this thesis is one of the products of the research.

Chapter I: INTRODUCTION

I.1. Development of Contact Lens Materials

The concept of a contact lens was described by Leonardo da Vinci about 500 years ago, but it was not until 1887 that the first contact lens was fitted into a patient's eye. Contact lenses were made of glass in about 1930, but because of brittleness and difficulties in processing and design, glass was later replaced by plastic, first in scleral form in 1938, and then in corneal form in 1948[3]. The pioneer plastic contact lens material is poly(methyl methacrylate) (or in short, PMMA), which is light, transparent, processable and non-toxic and which was commercially available at that time. Since then, the use of contact lenses has become widespread, and extensive clinical studies and fitting techniques on PMMA lenses were carried out to improve the performance of the lens. Despite the emergence of new contact lens materials in the past 13 years, PMMA contact lenses are still commercially available from manufacturers and worn by a majority of contact lens wearers.

In 1960, a new synthetic polymeric material called "hydrogel" was developed by O.Wichterle and D.Lim[4-6] in Czechoslovakia for a prosthetic implant material. Because of its biocompatibility and transparency, the hydrogel was also suggested for contact lens applications, and has made a big impact in the contact lens field in the past decade. This hydrogel material, which was claimed to be flexible and soft and compatible with the eye, is covalently cross-linked poly(2-hydroxyethyl methacrylate) (or in short, PHEMA). In 1965, the National Patent Development Corporation (NPDC) of the United States bought the exclusive right to this new material and gave the name "Hydron"

to the material and the lenses it manufactures. Further developments of the Czechoslovak lens were carried out in the United States since 1966 by the Bausch and Lomb Incorporated, and in the United Kingdom and Europe Since 1971 by Hydron-Lens Ltd., a joint venture of the NPDC and the Smith and Nephew Group. Structural variations of PHEMA contact lenses, designed to improve the properties of PHEMA hydrogel, include incorporation of an additional comonomer such as acrylamide[6-8], diethylene glycol methacrylate[9] and alkyl acrylate or methacrylate[10]. Besides these covalently cross-linked hydrogels, another type of hydrogel, which is an ionically cross-linked polymer of polyelectrolyte complexes of poly(vinylbenzyltrimethylammonium chloride) and polystyrenesulphonate, has been claimed to be more compatible with the living tissue and also suggested as a contact lens material[11]. This was not patented until 1971[12].

Along with the development of the hydrogel contact lens, another material called silicone rubber[poly(dimethyl siloxane)][13,14] was claimed in 1966 to be suitable for making flexible contact lens of high oxygen permeability.

So by September 1970, only three types of contact lenses were known, namely, PMMA, PHEMA and silicone rubber. But in the past three years, several patents on new contact lens materials appeared in the literature. Contact lenses made of soft thermoplastic materials such as polythene[15] and poly(vinyl chloride)[16] were patented in 1969, followed, in 1970, by a rigid thermoplastic material poly(4-methyl pent-1-ene)[17,18] which was claimed to be rigid in the eye and give good optical properties and high gas permeabilities. Besides silicone rubber, other elastomeric materials such as polybutadiene and ethylene propylene terpolymers were also patented as contact lens materials in 1971[19]. Later,a copolymer

of perfluoro-2-methylene-4-methyl-1,3-dioxolane and tetrafluoroethylene and a terpolymer of hexafluoroacetone, tetrafluoroethylene and ethylene were both claimed[20] to be suitable for making contact lenses of low refractive index, approaching that of the tear fluid. In the case of hydrogel contact lens materials, copolymers of N-vinyl lactams and alkyl acrylates[21], methyl methacrylate and acrylic acid[22], and polyvinyllactam and hydroxyethyl methacrylate[23,24] were all patented after 1970.

An entirely new approach to the contact lens materials has been described by Soehnges[25] in Germany, who used the gel of proteins extracted from mammalian eyes to manufacture contact lenses, but very little literature has been found with regard to this material.

In conclusion, polymeric materials which have been suggested in contact lens applications can be classified into four groups, namely,

(1) hydrogels such as covalently cross-linked PHEMA and ionically cross-linked polyelectrolyte complexes,

(2) thermoplastics such as PMMA, polyethylene, poly(vinyl chloride), poly(4-methyl pent-1-ene) and copolymers or terpolymers of tetrafluoroethylene,

(3) elastomers such as silicone rubber and polybutadiene, and

(4) biological polymers such as proteins from mammalian eyes.

Although numerous patents on the various types of contact lens materials appeared in the past ten years, most attention has been paid to the development of hydrogel contact lenses. By 1973, many types of hydrogel contact lenses were commercially available in England, and they are shown in Table I.1.

Table I.1.

Commercially available hydrogel contact lenses in England (by 1973)

Commercial name of the contact lens	Manufacturer	Commercial name of the material	Chemical constitutions of the material	Mode of manufacture
Soflens +	Bausch & Lomb Inc.	Polymacon	PHEMA	Spin-cast
Hydron	Hydron Lens Ltd.	Hydron	PHEMA	Lathe-cut
Naturalens		Bionite	HEMA-VP graft copolymer*	Lathe-cut
Sauflon	Contact Lens Manufacturing		MMA-VP copolymer*	Lathe-cut

+ "Soflens" contact lens has been approved by the Food and Drug Administration (FDA) of the United States in 1971 to be marketed in America.

* The chemical constitutions are not announced by their manufacturers, but they are thought to be so[26].

I.2. Methods of Contact Lens Manufacture.

As the material constituting a contact lens is either plastic or rubber, most of the conventional forming techniques used in the plastics and rubber industries can also be applied to manufacture some types of contact lenses. Because of the smallness of a contact lens, some special manufacturing processes (e.g. lathe-cutting and spin-casting techniques) have also been used, quite satisfactorily, particularly on hydrogel contact lenses.

According to the nature of the contact lens material as classified in the last section, different lens forming techniques have been

adopted by different contact lens manufacturers (as shown in Table I). However, the following methods have been used in general.

(1) Hydrogel Contact Lenses

Like other thermosetting plastic materials, a covalently cross-linked hydrogel such as PHEMA cannot be processed by heat and pressure in its dehydrated state (called "xerogel"). Methods which have been used to form covalently cross-linked hydrogel contact lenses are (a) lathe-cutting from a blank of xerogel[27,28], and (b) spin-casting of a polymerising monomer mixture in a lens mould in the presence[29] or absence[30,31,32] of a solvent (e.g. ethylene glycol) so that cross-linking takes place in-situ during polymerisation.

On the other hand, an ionically cross-linked hydrogel behaves like a thermoplastic. Gels of poly(vinylbenzyltrimethylammonium chloride) and poly(styrenesulphonate) have been plasticised with polar liquids (e.g. glycerine) and moulded into contact lenses under heat and pressure[12].

(2) Thermoplastic Contact Lenses

Thermoplastics are polymers which can flow under heat and pressure, and so thermoplastic contact lenses can be formed from the material by (a) pressing from a blank or lens-shaped disc, (b) moulding from a fine powder or pellet, (c) injection moulding (i.e. forcing a molten polymer into a lens-shaped cavity), and (d) lathe-cutting from a blank.

(3) Elastomeric Contact Lenses

Like moulding other rubber articles, an elastomeric contact lens is manufactured by vulcanizing the elastomer with a cross-linking agent (e.g. peroxide) in a lens mould under heat and pressure[19].

(4) Biological Polymer Contact lenses

Proteins are extracted from mammalian eyes and polymerised by

pressure or electrodialysis in a lens mould to form a contact lens[25]. However full experimental techniques have not been published.

In general, contact lens moulds are made of metal or glass. For thermoplastic and elastomeric contact lenses, the size and curvature of the mould and the thermal coefficient of the contact lens material all control the final size, shape and power of the lens. In the case of spin-cast hydrogel contact lenses, some additional variables such as speed of the rotating mould and the viscosity and amount of the polymerisation mixture also affect the size, shape and power of the lens. Also, the swelling factor of a hydrogel material has to be considered, whether in lathe-cut or spin-cast lenses, in order to give a hydrogel contact lens of required shape, size and optical power. A method[33] has been suggested to smooth the surface of a hydrogel contact lens or make it aspherical by heating the lens to above its glass transition temperature, shaping it and then cooling it down to room temperature at which the surface of the lens is polished to any required curvature. The lens is then returned to its original state by heating it above its glass transition temperature or swelling it in water.

Contact lenses removed from the mould are edged in order to give a comfortable wear in the eye. Hydrogel contact lenses are edged with a polishing paste in their dehydrated state, whereas elastomeric contact lenses cannot be edged or polished due to the rubbery nature of the material. It has been claimed[18] that the edge of poly(4-methyl pent-1-ene) contact lens can be made hydrophilic and soft by polymerising a water-swellable polymer (e.g. polyacrylic acid) onto a poly(4-methyl pent-1-ene) rod before it is lathe-cut or pressed into a contact lens.

Contact lenses made of hydrophobic materials such as PMMA, poly(4-methyl pent-1-ene) and silicone rubber are rendered hydrophilic by surface treatment or with a wetting solution. For example, PMMA

contact lenses have been made hydrophilic by free radical treatment[34,35] or by applying a coating of silica dioxide or titanium dioxide to the lens[36,37]. Silicone rubber lenses have been made wettable by treating the lens with tetraisopropyl titanate[38] or titanium tetra-alkoxide[39], or with a sulphonating agent and then a base[40], or with a scouring solution containing phosphoric acid and oxides of chromium[41].

The finished hydrogel contact lens is extracted in saline solution to remove any water soluble materials such as unreacted monomer, catalyst or degradation products during processing (if any). The hydrogel contact lens is then stored in saline solution after extraction, whereas thermoplastic and rubber contact lenses are stored in a soaking solution.

Because of high precision techniques involved in the manufacture of such a small article (i.e. contact lens), reproducibility of contact lenses has not been satisfactory, particularly hydrogel lenses where swelling factors and isotropy of swelling affect the final lens shape, size and optics.

Internal stresses have been reported[42] in those contact lenses manufactured from processes involving pressure and heat. The internal stresses result from the orientation of polymer chains during flow of the material in the mould, and will cause the lens to change shape in a preferred direction during any subsequent relaxation process. But the stresses can be largely removed by annealing the lens in its mould at a sufficiently high temperature. Lathe-cut lenses, on the other hand, are more likely subjected to surface strains resulting from the shear force during cutting and bulk deformation due to mounting. However all the stresses can be minimised with extreme care and skill during manufacture.

I.3. Contact Lenses and the Eye

I.3.A. Function of a Contact Lens

A contact lens is normally prescribed to patients with poor visions and in some cases, for therapeutic application. In addition, because of an unique physical property of hydrogels, hydrogel contact lenses have been examined for medication application, i.e. as a vehicle for drugs. However the main application of a contact lens is to correct vision defects such as myopia (i.e. short-sightedness), hypermetropia (i.e. long-sightedness) and astigmatism.

In a normal eye, a ray of light from an object diffracts through the cornea and the lens of the eye and falls on the retina (See Figure I.1.). When the eye is abnormal in vision, the object image falls in front of the retina (i.e. short-sightedness) or behind the retina (i.e. longsightedness) or on the retina but different planes (i.e. astigmatism). The function of a contact lens, its design and fitting techniques are to bring the object image to fall on the retina of the eye wearing the contact lens, as in the case of a normal eye.

I.3.B. Structure of the Cornea

The cornea of the eye is a vital part because the cornea is the major refractive surface of the eye. The cornea consists of five layers, namely (from the external surface), epithelium, Bowman's membrane, stroma, Descemet's membrane and endothelium (see Figure I.2.). Because of thinness of Bowman's membrane and Descemet's membrane, and their respective similarities with stroma and endothelium, both Bowman's membrane and stroma, and Descemet's membrane and endothelium can be grouped together. As a result, for simplicity, a cornea can be considered to consist of essentially three layers, i.e. epithelium

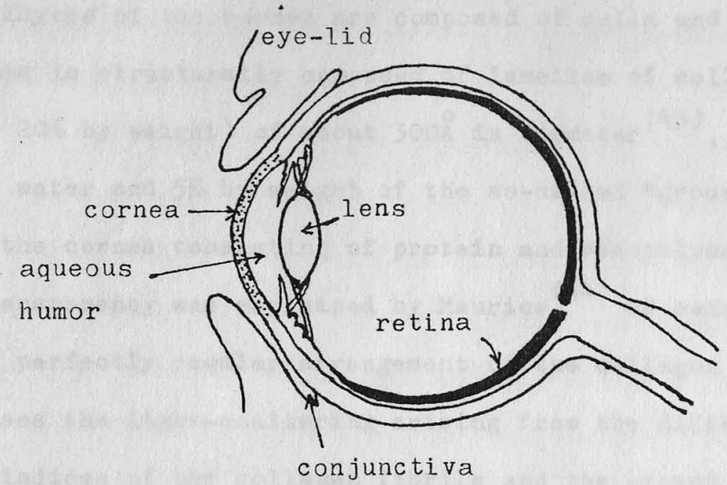


Figure I.1 The structure of the eye.

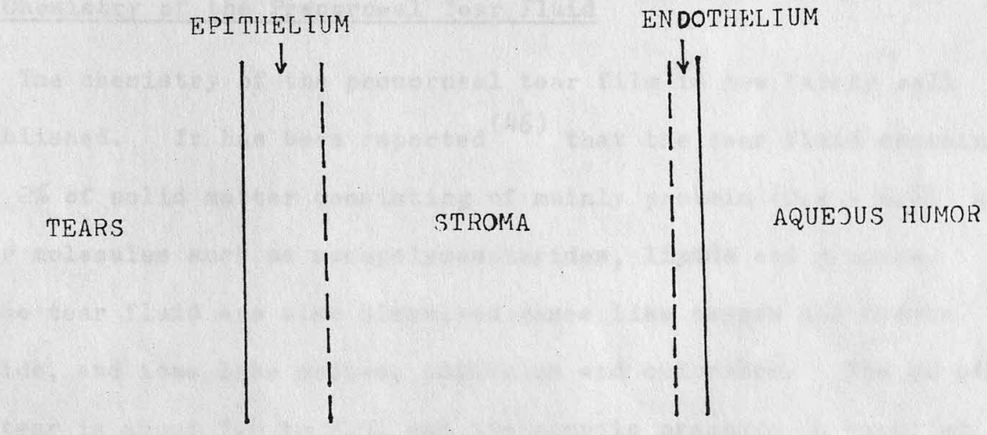


Figure I.2 Schematic arrangement of corneal layers (Layer thicknesses are not scale).

(0.04 mm thick) stroma (0.45 mm thick) and endothelium (0.01 mm thick). All the three layers of the cornea are composed of cells and in addition, stroma is structurally composed of lamellae of collagen fibrils (about 20% by weight) of about 300Å in diameter[43], embedded in a matrix of water and 5% by weight of the so-called "ground substance" of the cornea consisting of protein and mucopolysaccharides. The corneal transparency was explained by Maurice[44] as being a consequence of perfectly regular arrangement of the collagen fibrils, which neutralises the light-scattering arising from the difference in refractive indices of the collagen fibrils and the ground substance. Besides collagen, soluble protein and mucopolysaccharides, the cornea contains about 75% by weight of water (one-fifth of which can be expressed from the tissue and the rest can be removed only by drying[45]) enzymes, amino-acids, glycogen, lipids, cations (e.g. Na^+, K^+, Ca^{2+}) and anions (e.g. lactate, chloride, sulphate, phosphate). The cornea is separated, on one side, from the lens of the eye (see Figure I.1) by the aqueous humor, and on the other side, from the atmospheric air by a layer of tear fluid (about 10μ thick).

I.3.C. Chemistry of the Precorneal Tear Fluid

The chemistry of the precorneal tear film is now fairly well established. It has been reported[46] that the tear fluid contains 1 or 2% of solid matter consisting of mainly protein (0.2 - 0.6%) and other molecules such as mucopolysaccharides, lipids and glucose. In the tear fluid are also dissolved gases like oxygen and carbon dioxide, and ions like sodium, potassium and chlorides. The pH of the tear is about 7.4 to 7.5, and its osmotic pressure is reported to be roughly the same as that of plasma.

It has long been known that a continuous tear film covers the

corneal epithelium, giving a good optical surface of the cornea. In fact, the epithelial surface of the cornea has been found to be irregular and has a critical surface tension of about 28 dynes/cm[46] or 31 dynes/cm[47] deduced from contact angle measurements of the cornea. Although the surface tension of pure water is known to be 72 dynes/cm., that of the pre-corneal tear film has been recently found to be about 46 dynes/cm[48], due to the presence of the dissolved materials in the tear fluid. As it is a known fact that a liquid which has a surface tension greater than the critical surface tension of a solid will not wet it[49], the tear fluid was not expected to spread and wet the epithelium. The wetting of the corneal epithelium by the tear fluid in the eye was explained[47,50] by the adsorption of mucin (about 0.02 - 0.4μ thick), produced by the conjunctival goblet cells and spread by the eye-lids, on the epithelial surface. This raises its critical surface tension, presumably to $>$ 46dynes/cm. (the surface tension of the tear fluid). The evaporation of the tear fluid in the eye is reduced by a superficial layer of lipid (about 0.004 - 0.4μ thick) produced by glands of the eye-lid. The existence of the superficial lipid layer on the tear fluid has recently been demonstrated[51]. As the tear fluid thins due to evaporation, the lipid layer eventually contaminates the adsorbed mucin layer on the epithelial surface, producing a hydrophobic surface. Blinking is therefore thought to be essential to resurface the mucin layer, and also to respread the lipid layer.

I.3.D. Oxygen Utilization and Consumption of the Cornea, and Critical Oxygen Tension at the Corneal Surface

It has been accepted from experimental evidences that carbohydrate breakdown in the corneal epithelium is the principal source of energy

required for the maintenance of corneal clarity. The energy is utilized to rid the tissue of water so that it is retained in its partially dehydrated "deturgescent" state and the stromal collagen fibrils are maintained in an orderly and regular arrangement embedded in the mucoid material. If the cornea is deprived of its oxygen supply, water accumulates within the cells and leaks out into the stroma and swells the gel mucoid matrix, disrupting the spatial regularity of the collagen fibrils. As a result, the corneal transparency decreases and the cornea becomes thick, and this phenomenon is called "corneal edema".

Both Smelser[52] and Langham[53] suggested that the human cornea utilized oxygen for its metabolic requirements from the atmosphere, and this was supported by Hill and Fatt[54,55,56]. With the recent development of ultramicro oxygen electrode which enables the oxygen tension to be measured within the cornea, it has been proved experimentally[57] that the oxygen supply to the corneal epithelium and stroma of rabbit eyes is mainly through the utilization of atmospheric oxygen. It is reckoned[58] that the atmospheric oxygen is carried to the cornea via four pathways, i.e. (a) by air that is dissolved in the tears which oxygenates the cornea by a continuous bathing process with blinking, (b) by the aqueous humor, (c) by the blood of limbal capillaries, and (d) by the blood of the conjunctiva vessels. All the existing evidences suggest that the pathway (a) is most important.

Making use of the knowledge and assumptions of thickness, oxygen and carbon dioxide transmissibilities of each of the epithelium, stroma and endothelium, Fatt et al calculated the steady-state distribution of oxygen and carbon dioxide in the in-vivo cornea of (i) an open eye in the air[59], (ii) a closed eye[59], and (iii) an open

eye in nitrogen atmosphere[60]. They found that in the open eye[59] under normal atmospheric conditions, oxygen diffused across the epithelium into the cornea from the tear fluid, and out through the endothelium into the aqueous humor, whereas carbon dioxide (a metabolic product) moved from the aqueous humor through the cornea and out into the air. For the closed eye[59], about two-thirds of the oxygen required moves into the cornea across the epithelial surface from the capillaries in the palpebral conjunctiva and the remainder across the endothelial surface from the aqueous humor, whereas carbon dioxide moves across both surfaces and out into aqueous humor and palpebral conjunctiva. Fatt[60] also found that for an open eye in a nitrogen atmosphere, the cornea obtained oxygen from the aqueous humor, but the stroma had severely reduced oxygen tension and the epithelium was almost anoxic, and all the carbon dioxide discharged into the aqueous humor.

The oxygen permeabilities of the stroma[62], epithelium and endothelium[63] have been reported as 300×10^{-10}, 190×10^{-10} 54×10^{-10} cc.mm.$(cm^2$ tissue$)^{-1}$.sec^{-1}.cm Hg^{-1} respectively.

By measuring the oxygen uptake from a limited reservoir of a Clark type polarographic electrode by "in vivo" human cornea, Hill and Fatt[54] calculated the oxygen flux across the cornea from the air as 4.8 to 7.2 µl cm^{-2}.hr^{-1}, normally taken as 5 µl.cm^{-2}.$hr.^{-1}$. Later Fatt et al[61] found that the total oxygen flux (from the anterior epithelial surface and/or posterior endothelial surface) into the cornea depended on the oxygen tension at the epithelial anterior surface between 0 and 20 mm Hg, but remained constant at a value of about 8.5 µl.cm^{-2}.hr^{-1} at any oxygen tension beyond 20 mm Hg. (See Figure I.3).

The oxygen consumption rates of the epithelium, stroma and

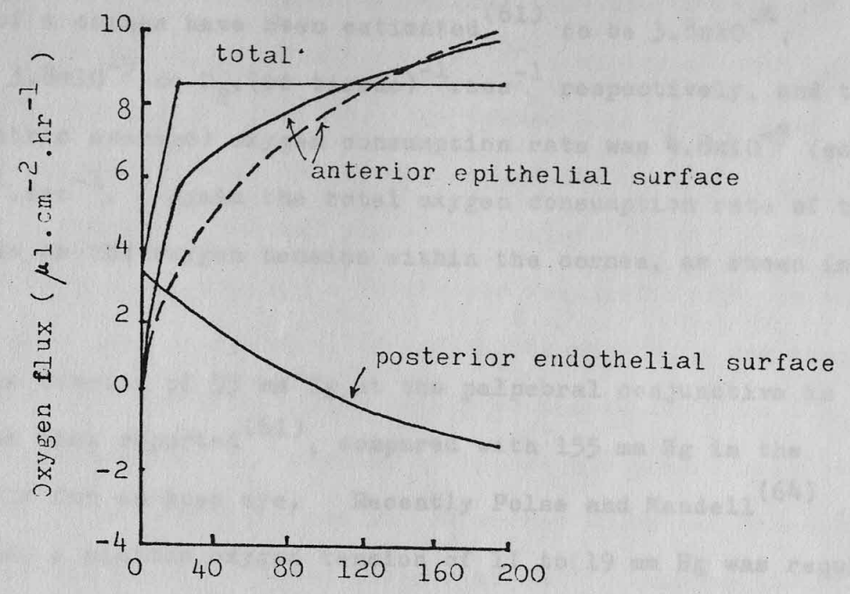


Figure I.3 Oxygen flux into the surfaces of the cornea as a function of the oxygen tension at the epithelial anterior surface [61]. —— calculated ; ---- experimental

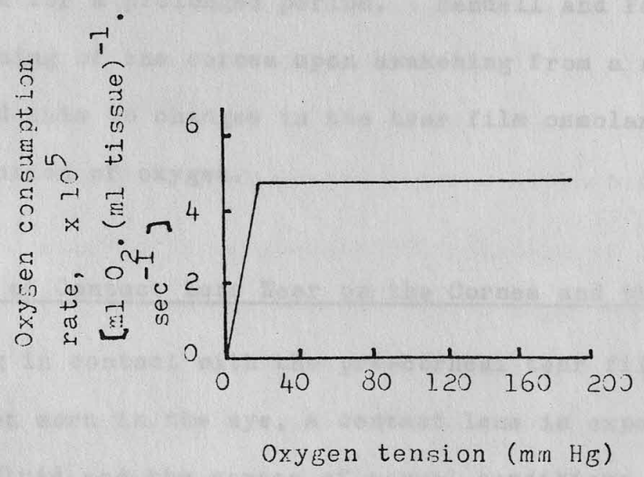


Figure I.4 Oxygen consumption rate as a function of oxygen tension for a single layer cornea. The rate is the volumetric average of the rates in the epithelium, stroma and endothelium [61].

endothelium of a cornea have been estimated[61] to be 3.8×10^{-4}, 1.1×10^{-5} and 3.8×10^{-5} cc O_2.(cc tissue)$^{-1}$.sec^{-1} respectively, and the total (volumetric average) oxygen consumption rate was 4.8×10^{-5} (cc O_2).(cc tissue)$^{-1}$.sec^{-1}. Again the total oxygen consumption rate of the cornea depends on the oxygen tension within the cornea, as shown in Figure I.4.

An oxygen tension of 55 mm Hg at the palpebral conjunctiva in a closed eye has been reported[61], compared with 155 mm Hg in the atmospheric air for an open eye. Recently Polse and Mandell[64] determined that a minimum oxygen tension of 11 to 19 mm Hg was required at the corneal epithelial surface in order that the cornea showed no sign of swelling. They concluded that the epithelium requires only a fraction of the oxygen which is normally available, i.e. about one-tenth of that available from the atmosphere when the eye is opened, and about one-fourth of that available from the palpebral conjunctiva when the eye is closed.

However corneal swelling has been observed even when the eyes are closed for a prolonged period. Mandell and Fatt[65] found a 3 to 4% thickening of the cornea upon awakening from a night's sleep, and attributed this to changes in the tear film osmolarity, as there was no deprivation of oxygen.

I.3.E. Effect of Contact Lens Wear on the Cornea and the Tear Fluid

Being in contact with the pre-corneal tear film and close to the cornea when worn in the eye, a contact lens is expected to affect both the tear fluid and the cornea of normal conditions.

Since the cornea utilises oxygen from the atmospheric air dissolved in the pre-corneal tear fluid, a contact lens when worn in the eye will serve as a barrier to the access of the atmospheric oxygen

to the epithelium. The influence of contact lens wear on the cornea was first reported in 1952 by Smelser et al[52], who found that after hours of wearing PMMA contact lenses, the cornea lost its optical properties by increasing haze in the vision and halos about light sources (i.e. symptons of corneal edema). Smelser[66] and Kinsey[67] both reasoned that the atmospheric oxygen was required by the cornea for its metabolism and maintenance of corneal transparency and that the contact lens cut down the oxygen supply to the cornea from the atmosphere and slowed down the removal of hypertonic pre-corneal tear film, resulting in an overhydration of the cornea which was the cause of corneal haze and halos. Later Smelser et al[68] studied the effect of PMMA contact lenses on the corneas of guinea pigs, and observed a rise in the level of lactic acid in the cornea, a consequence of insufficiency in corneal oxygen supply, as explained by Langham[53]. It was thus concluded that when the cornea suffered from a lack of oxygen, glucose was broken down increasingly to lactic acid, which yields less energy than glucose oxidation[69], and glycogen was used up in the epithelium cells[70]. Other observations of oxygen deprivation in the cornea include an increase in lactate concentration[71] and a decrease of trans-corneal potential[57]. A direct approach was made by Hill et al[56] who found, using a Clark type electrode, a depletion of oxygen in the cornea after a contact lens wear.

In the study of steady-state distribution of oxygen and carbon dioxide in the vivo cornea covered with a tight PMMA contact lens, Fatt et al[61] found that under this condition, the cornea obtained oxygen from the aqueous humor but the epithelium was deprived of oxygen due to its distance from the aqueous humor, and that all the carbon dioxide was discharged into the aqueous humor.

The production rates of carbon dioxide in the cornea has been assumed[61] to be the same as the oxygen consumption rate, i.e. 4.8×10^{-5}

(cc CO_2).(cc tissue)$^{-1}$.sec^{-1}. Fatt et al[61] found that even an impermeable contact lens did not raise the carbon dioxide tension in the cornea above that of a closed eye. The fact that the closed eye (e.g. during sleep) seems not to be damaged by the high carbon dioxide level leads to the belief that any contact lens material (of high or low carbon dioxide permeability) will not raise the carbon dioxide tension in the cornea to such an extent to damage the cornea, whether the eye is opened or closed. Fatt et al[61] reasoned this being due to the transmissibility for carbon dioxide of the cornea being twenty-fold greater than oxygen, and carbon dioxide/transmissibility of most polymers being five-fold greater than oxygen transmissibility.

The effect of contact lens wear on the pre-corneal tear fluid was not paid attention to until recently, although it was mentioned by Smelser et al[52] twenty years ago. It was recently found that during adaptation to corneal contact lenses, (i) the total protein content of the tears was frequently decreased[72], (ii) the sodium and potassium concentrations of the tears were changed[73], and (iii) there were significant hypertonic and hypotonic variations in the osmotic pressure of the tears, which subsequently stabilised at a similar value to that which existed before fitting[74].

Recent studies by Mandell et al[75,76] showed that during adaptation to contact lens wear, the cornea thickened between 2 and 4%, and attributed this to the increased lacrimation associated with the adaptation process. They also observed that tightly-fitting contact lenses deprived the cornea of oxygen and further increased the corneal swelling.

I.4. Clinical Comparisons of Poly(2-hydroxyethyl methacrylate) hydrogel, Poly(methyl methacrylate) and Silicone Rubber Contact Lenses.

As mentioned earlier three types of contact lenses were known in

1970, i.e. PHEMA, PMMA and silicone rubber contact lenses. Since the introduction of PHEMA hydrogel contact lenses, the terms "hard" and "soft" have been adopted and referred to PMMA lenses and PHEMA hydrogel lenses (and other hydrogel contact lenses) respectively, purely because of extreme difference in physical feelings with fingers during handling. Because of the emergence of other new materials during the development of contact lens materials, it has been recently suggested[77] that "hard" and "soft" may be obsolete, and that the terms "rigid" and "flexible" may be more appropriate to describe all the contact lenses. This means that under the latter classification, PMMA and poly(4-methyl pent-1-ene) are rigid contact lenses, while PHEMA and silicone rubber are flexible contact lenses.

Ever since hydrogel contact lenses were introduced, clinical work has been very active on comparing PHEMA hydrogel contact lens with PMMA lens and later with silicone rubber lens. Although hydrogels of polyelectrolyte complexes have been suggested in contact lens applications no clinical work has yet been found on this material. Some physiological aspects of the cornea were studied and understood. For the past two to three years, because of the emergence of various types of hydrogel contact lenses, clinical works have been concentrated on evaluating various hydrogel lenses in the eye, and on the micro-biological and pharmocological aspects of hydrogen contact lenses.

However, in order to understand the advantages and disadvantages of these three types of contact lenses, some of the clinical works which have been carried out on them, particularly their comparisons, are described as follows.

I.4.A. Comfort and Vision of Contact Lens Wear

Comfort and visual acuity are two of the many criteria in contact lens wear. It has been reported that a PMMA lens gives a perfect and stable vision because it is not affected by the eye-lid pressure

during blinking, and corrects astigmatism quite well because of a stable lacrimal "lens" formed between the contact lens and the cornea. But on the other hand, a PMMA lens exerts mechanical pressure on the cornea and wearers can sometimes feel a foreign body in their eyes particularly the edge of the lens, and so a PMMA lens is uncomfortable to wear. A PHEMA contact lens was reported to rarely give a perfect vision, and correct astigmatism only to a lesser degree, but it is comfortable to wear because of the softness, it is said, of the material. However, the prime reason of vision fluctuation in a hydrogel contact lens is thought to be the fit of the lens. A properly fitted hydrogel contact lens gives a more stable vision. The shape, size and optics of a hydrogel contact lens are affected by many factors such as ambient temperature, relative humidity, rate of blinking and tearing, and nature and concentration of solution bathing the lens. Hydrogel contact lenses are not recommended to poor blinkers nor are they normally recommended to be worn while swimming. It has been found[78,79,80] that a PMMA lens affects corneal sensitivity while a hydrogel lens has little effect. Wearing a PMMA lens has been reported[81] to reduce the rate of blinking whereas wearing a PHEMA lens has little effect on the blink rate, and this was attributed to the foreign body effect in the hard lens wear. Clinical work on silicone rubber lenses has recently been reported[77,82]. A silicone rubber lens does not give as good vision as a PMMA lens, and on the other hand, is not as comfortable as a PHEMA lens, but is able to correct astigmatism to some degree because of possession of rigidity and elasticity.

It has now been accepted that fitting a PMMA lens requires rather exacting handling operations so that the lens centres in the eye and

floats in the tear fluid. Fitting a PHEMA lens is comparatively easy and does not usually need all the sophisticated fitting equipment used in the case of a PMMA lens. Like fitting a PMMA lens, a hydrogel contact lens may be fitted to curvatures and diameters. While a PMMA lens is used in corneal form, a PHEMA lens is used in semi-scleral form. A silicone rubber lens is fitted and the fit is checked just as for a PMMA lens. Adaptation time of wearers to a PMMA lens sometimes takes as long as several weeks, but in the case of a PHEMA lens, wearers were reported in some cases to wear it for eight to ten hours in the first day. The initial physical comfort of a silicone rubber lens is comparable to a well-fitted PMMA lens, but the adaptation time is much shorter (normally 1 week). However the adaptation time of a lens varies from one patient to another.

It is understood that a PMMA lens floats in the tear fluid in the eye. The existence of a continuous pre-corneal tear film and its surface tension on a contact lens were thought [83,84] to centre the contact lens at the cornea. In order to be physically compatible with the tears, a contact lens has to be wettable by the tear fluid. A PMMA contact lens or a silicone rubber lens is normally applied with a wetting solution before it is inserted into the eye, so that when it is worn there is a continuous tear layer on the lens, and the vision and comfort will not be impaired. On the contrary, a PHEMA contact lens is known to be wettable by the tear fluid in the eye because no wetting solution is required in practice. However a PHEMA lens has to be soaked in a solution isotonic with the tear fluid. Hypotonicity or hypertonicity of the lens will cause stinging sensation to the eye.

I.4.B. Physiological Aspects of Contact Lens Wear

A good contact lens must not only give comfort and stable vision, but also must not interrupt physiological behaviours of the cornea. As the cornea obtains oxygen from the pre-corneal tear fluid, a contact lens worn in the eye is expected to interrupt the access of oxygen to the epithelial surface.

A comparison of the effects of contact lenses made of PHEMA (Geltakt : Czech. lens), silicone rubber and PMMA on lactic acid level, hydration and transparency of the cornea was made by Praus et al[70], who applied these three types of contact lenses of equal size (large enough to cover completely the cornea so as to limit oxygen transmission to the cornea by diffusion through the lens only) to rabbit corneas for eight hours. They concluded that a PHEMA lens caused less oxygen deficiency than a PMMA lens and did not differ much from a silicone rubber lens in allowing oxygen access to the cornea under the same experimental conditions. Their results are consistent with Czechoslovak workers' earlier clinical study[85] and laboratory measurements[86] that PHEMA hydrogel has a considerably greater oxygen diffusion coefficient than PMMA. But the Czech's clinical results are contradicted by the findings of the American workers. Thus Takahashi et al[87] measured the oxygen transmissibilities of both PHEMA and PMMA contact lenses and found them to be about the same, no higher than 4×10^{-10} cc (S.T.P.). $mm.cm^{-2}.sec^{-1}.$ cm Hg^{-1}, and Hill et al[8] found that a silicone rubber contact lens was very much permeable to oxygen than a PHEMA contact lens.

Further to this, Hill[89,90] studied the effect of these three types of contact lenses on the corneal respiration of rabbits under static conditions. By studying the rate of exhaustion of the oxygen

reservoir at the tip of a polarographic probe at the cornea after 90 second periods of no lens wear and applications of a hydrogel lens, a PMMA lens and a silicone rubber lens, Hill found that under these conditions, both PHEMA and PMMA contact lenses resulted in a greater oxygen deprivation (both had an oxygen uptake rate 2.4 times faster than in the case of no lens wear), and that the oxygen uptake rate in the case of silicone rubber lens was statistically indistinguishable from the normal uptake rate of the no-lens eye. This means that insignificant amount of oxygen permeates through both the PHEMA and PMMA lenses at the same degree to meet the normal requirements of the cornea, while a silicone rubber lens has little interference on the corneal respiration.

Recently Burns et al[71] found that the glycogen in a rabbit corneal epithelium decreased and the lactate increased from its original level in hours of wear of a PHEMA lens (Geltakt) or a PMMA lens, and that there was no significant difference of the epithelium glycogen, glucose, adenosine triphosphate and lactate levels during silicone /rubber lens wear from those of a normal cornea. They concluded that both PHEMA and PMMA contact lenses deprived the cornea of oxygen, while a silicone rubber lens permeated sufficient oxygen from the atmosphere for normal metabolism of the cornea.

Hamano et al[92] also found that wearing a hydrogel contact lens increased the corneal thickness and decreased both the glycogen content and the trans-corneal potential of the cornea to a lesser extent than wearing a PMMA contact lens. Later, using an oxygen microelectrode to measure the oxygen tension in the corneal stroma with various contact lenses placed "in-situ", Hamano et al[57] also found that the oxygen tension dropped to zero at a slower rate on wearing a hydrogel contact

lens than a PMMA lens, and that the oxygen tension was not affected on wearing a silicone rubber contact lens.

Although some of the results described above have been controversial, it becomes apparent, as more and more work has been carried out, that a PHEMA contact lens has less effect than a PMMA lens on the corneal respiration during the contact lens wear, and a silicone rubber contact lens has little effect on the corneal respiration.

Despite the observation that both the PHEMA and PMMA contact lenses are very much less permeable to oxygen than a silicone rubber lens, the fact that both the lenses have been accepted for daily wear and that corneal edema are not always reported on the wearers both seem to suggest that access of oxygen from the atmosphere to the cornea is possible by another route in addition to diffusion through the lenses. It has been accepted that fresh tear fluid between a contact lens and the corneal epithelium plays an important role in transporting the atmospheric oxygen to the epithelium for its physiological needs. There are different mechanisms suggested for the refreshment of this tear layer. It is believed that as a PMMA contact lens floats in the tear fluid in the eye, the "rocking" (up and down movement) of the lens during blinks helps the circulation of the tear fluid layer[77]. In the case of hydrogel contact lens wear, it is reckoned[77] that it is the rippling of the lens during blinks that replenishes the tear layer under the lens. In fact Hill suggested earlier[90] that adequate tear exchange by the lid and eye movements would be required for a hydrogel contact lens as in the case of a conventional PMMA lens. A tight hydrogel contact lens (i.e. absence of the tear layer) has been found to deprive the cornea of oxygen very soon[76]. In the case of silicone

rubber lens wear, the "rocking" (up and down movement) and "bellowing" (in and out movement) of the lens were thought[82] to renew the tear fluid under the lens. However the high oxygen diffusion through a silicone rubber lens also helps to build up the oxygen tension at the epithelium - tear interface for normal physiological requirements of the cornea.

Although the tear exchange brings fresh oxygen to the cornea, the tear exchange is never as complete in contact lens wear as it is without a lens. It is also believed[58] that with a PMMA lens in the eye, the high absorption of infra-red radiation by the polymeric material increases the temperatures of the tear fluid and of the epithelial cells, resulting respectively in a reduced solubility of oxygen in the tears and an increased metabolic rate of the epithelium. The heat-up of the tears has been observed clinically as an appearance of tiny bubbles of gas in the apical tear pool. All these effects contribute to the oxygen insufficiency at the cornea and the occurrence of corneal edema in some PMMA contact lens wearers. The fact that a hydrogel contact lens is well tolerated by people whose corneas are unable to tolerate the hypoxic situation created by a PMMA lens leads Feldman[58] to suggest that a hydrogel having a high water content warms the cornea to a lesser degree if there is any effect by the radiation absorption. However a PHEMA hydrogel contact lens has been found[93] to have a thermal conductivity of 6.6×10^{-4} $cal.cm^{-1}.sec^{-1}.{}^{o}C^{-1}$, comparable to that of the cornea, and higher than that of a PMMA contact lens (4.4×10^{-4} $cal.cm^{-1}.sec^{-1}.{}^{o}C^{-1}$.).

As a PMMA contact lens is impermeable to oxygen, attempts to improve the oxygen diffusion through the lens have been carried out by perforating up to 1.0 mm in diameter in the lens[94], but this

approach was not successful because no significant improvement in oxygen permeation was achieved.

The critical oxygen tension required by the cornea has been found to be 11 to 19 mm Hg[64]. Independently, Fatt and Hill[95] showed that blinking rocked a contact lens sufficiently to maintain an oxygen tension of about 20 mm Hg under the lens. Later Jauregui and Fatt[96] used a micro-oxygen sensor to measure the oxygen tension under a contact lens while it was worn in a completely normal manner, and found that under a tight fitting contact lens the oxygen tension was close to zero whereas under a well-fitted contact lens the oxygen tension was about 20 mmHg. Since clinical work has shown that a well-fitted contact lens does not normally cause swelling in the cornea, it follows that Fatt etal's result supports quite well the findings of Polse et al.

It is understood that refreshment of the tear pool between a contact lens and the cornea is essential for oxygen supply to the corneal epithelium. In fact, Hill[97] has even used a contact lens with a base curve flatter than the cornea to allow oxygen movement to the cornea. However, it has been found[95] that an increase in fractional tear replacement is more effective than increasing blink rate or decreasing tear reservoir volume in increasing oxygen tension under a lens. The mixing efficiency of the process that replenishes the tears under the lens at each blink has been found[98] to be about 10%.

Besides dissolved gases, ions and larger molecules (such as proteins, glucose and mucopolysaccharides) are also contained in the tears. The permeabilities of three hydrogel contact lenses (Geltakt, Physiological Polymer and Viziblank, in order of increasing water

content) to radioactive labelled water, D-glucose and L-amino acids have been studied recently, and the author[99] came to two major conclusions. (a) These hydrogel polymers had a minimum "pore" diameter of approximately 100 Å since glucose and amino acids could pass through them. (b) Only 1.3% of oxygen required for corneal metabolism could diffuse through these three hydrogel materials, because the diffusion of water through these materials during the first eight hours was found to be of the order of 1.3% of the total reservoir used. However their conclusions are not consistent with other workers' findings.

I.4.C. Microbiological and Pharmocological Aspects of Contact Lens Wear

Because of the contact of a contact lens with the cornea, hygiene of contact lens wear has to be considered. When not in use, PMMA and silicone rubber contact lenses are kept in a storage solution. Before the lens is to be inserted into an eye, it is cleansed and wetted with a suitable wetting solution (containing preservatives). Preservatives such as benzalkonium chloride, ethylene diaminetetra-acetic acid (EDTA) and chlorobutanol have been used in wetting solutions for PMMA contact lens[77]. These preservatives have been used experimentally in the wetting solution for silicone rubber contact lenses but they were found to cause discomfort during wear[77]. On the other hand, a hydrogel contact lens is kept in a normal saline solution which is isotonic with the tear fluid. Solutions containing the preservatives are found to be unsuitable for cleansing hydrogel contact lenses because the preservatives may be absorbed and later released in the eye during wear causing chemical burns. Other ways of sterilisation of hydrogel contact lenses have therefore become very

essential in keeping the lens free of bacteria which may otherwise infect the cornea when the lens is worn. Several ways of sterilising hydrogel contact lenses have been reported[100], such as (i) boiling the lens in saline, as recommended for Soflens contact lens; however boiling was found to kill pathogens and not spores, (ii) treating the lens with 3% hydrogen peroxide (pH 3) followed by neutralisation with sodium bicarbonate solution, as recommended for Bionite contact lens, and (iii) storing in an isotonic sterilising solution containing chlorohexidine gluconate or a mixture of thimerosal, EDTA and chlorohexidine. Before any sterilisation, the lens is normally cleansed to free of proteins otherwise proteins will be denatured by boiling or hydrogen peroxide treatment, and precipitate on the lens, or will deactivate the germicide.

The scanning electron microscope technique has been used (at magnifications of 15,000 and 36,000) to examine a hydrogel contact lens but failed to detect any "pores" in the lens, and so it was reckoned that bacteria being at least 700 times bigger in diameter than an oxygen molecule, cannot get through the "pores", but is adsorbed at the surface of the lens[101]. It is thought that various components of the tears may be absorbed into the lens through the "pores" and then serve as a reservoir of nutrient materials which could be utilised by the surface bacteria. Although lysozyme is present in the tears and reported to have bactericidal effects, it seems that it does not have much effect on the bacteria on the lens surface.

Mucus has been reported to accumulate on every one of the three types of contact lenses, and so the lens has to be removed from the eye and cleansed often after it has been worn for some time.

I.4.D. Summary of the Advantages and Disadvantages of the Three Types of Contact Lenses

By comparing the clinical performances of the three types of contact lenses, their advantages and disadvantages can thus be summarised in Table I.2.

It should not be difficult by now to point out that an ideal contact lens material should be at least hydrophilic, soft and flexible, and very permeable to oxygen, particularly the latest.

Table I.2.

Comparisons of properties of PMMA, PHEMA and silicone rubber contact lenses.

Properties	PMMA lens	PHEMA lens	Silicone rubber lens
Hydrophilicity	hydrophobic	hydrophilic *	hydrophobic
Hardness	hard	soft *	soft *
Rigidity	rigid	flexible *	flexible *
Permeability to oxygen	impermeable	some degree	very permeable *
Fitting	straightforward	simpler than PMMA *	as PMMA lens
Adaptation time	about 3 weeks	very short *	about 1 week
Comfort	less comfortable than silicone rubber	very comfort-able *	moderately comfortable
Visual acuity	very good *	not as good as PMMA	not as good as PMMA
Correction for astigmatism	for high astigmatism *	for low astigmatism	for low astigmatism
corneal edema	frequent	occasional	absent *
Sterilisation technique	simple *	complex	simple *
Storage between use	simple *	in saline solution	simple *
Resistance to mishandling	Good *	poor	good *
Life span of the lens	about 2 years *	1½ to 2 years	about 2 years
Cost (compared with PMMA lens	- *	more expensive	more expensive

Foot-note: Advantages are marked with *, and disadvantages otherwise

I.5. Literature Survey on the Chemistry and Properties of Poly(2-hydroxyethyl methacrylate) Hydrogel

Because of hydrophilicity and biocompatibility, synthetic cross-linked hydrogels seem to be the best class of material for contact lens applications, It is logical to use poly(2-hydroxyethyl methacrylate) hydrogel as a model as well as a stepping stone because it is the first synthetic hydrogel discovered for biomedical and contact lens applications, and its properties are most extensively studied. Therefore it is worth while to review the chemistry and properties of poly(2-hydroxyethyl methacrylate) hydrogel, with reference to contact lens applications wherever necessary, with an attempt to understand some of the basic properties of hydrogels in general.

In fact hydrogels are water-containing gels and include various chemical substances including gelatin, polysaccharides, polyelectrolyte complexes, cross-linked acrylamide polymers, cross-linked hydroxyl containing methacrylic (or acrylic) ester polymers, etc.[102] It is the latest class of hydrogel which is of particular interest.

Cross-linked hydrogels of PHEMA (and its copolymers) have been applied in the form of implants in almost all branches of surgery[103,104], in ophthalmology, particularly contact lens application, in cosmetic and wound-dressing film[105], in soft-lining of dentures[106], and in medication[107, 108, 109]. The hydrogels have also been applied in other fields such as desalination by reverse osmosis[110,111], gel permeation chromatography[112,113] and fibre application[114].

I.5.A. Chemical Structure and Microstructure

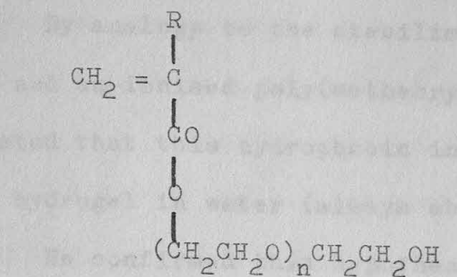
Poly(2-hydroxyethyl methacrylate) (II; i.e.I having $R=CH_3$, n=0 shown in Figure I.5) belongs to a class of polymers of methacrylic (or acrylic) esters containing at least one hydroxyl group in the side chain (general structure I shown in Figure I.5). The monomer HEMA can be prepared by transesterification of methyl methacrylate[115,116] or by the reaction of methacrylic acid with ethylene oxide[117,118].

Diethylene glycol methacrylate (i.e. I having $R=CH_3$, n=1), triethylene glycol methacrylate (i.e. I having $R=CH_3$, n=2) and 2-hydroxyethyl acrylate (i.e. I having R=H, n=0) do exist. The corresponding diesters particularly ethylene glycol dimethacrylate (EGDMA) (IV; i.e. III having n=1) are used to cr oss-link the polymer of HEMA to form a three-dimensional network. Other diesters known are diethylene glycol dimethacrylate (III having n=2), triethylene glycol dimethacrylate (III having n=3) and tetraethylene glycol dimethacrylate (III having n=4).

The cross-linked structure of PHEMA results in the polymer being solvent resistant, thermally stable, incapable of flow under heat and pressure, but able to swell in water and other solvents. These are important in contact lens applications where dimensional stability of a contact lens is required and where a spin-cast contact lens involves optionally the use of a solvent in the polymerisation mixture.

Refojo[119] postulated that PHEMA hydrogel has, in addition to its covalently cross-linked structure, a secondary structure stabilised by hydrophobic bondings, which may be localised between the α-methyl groups of the repeating units, or perhaps but less likely, distributed throughout the whole hydrophobic backbone of the

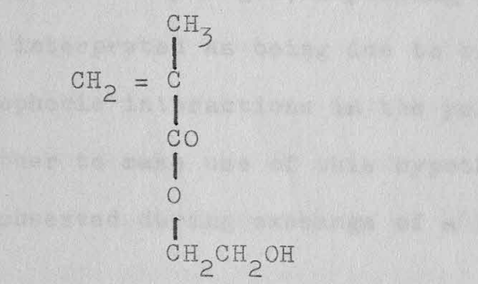
(I)



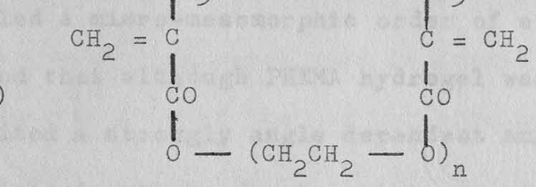
(R=H, CH_3)

(n=0, 1, 2, ...)

(II)



(III)



(IV)

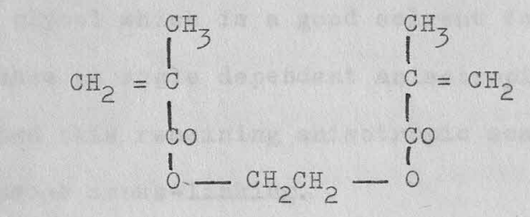


Figure I.5.

Chemical structure of glycol methacrylates/acrylates and glycol dimethacrylates.

network. By analogy to the stabilisation of structures of globular proteins and un-ionised poly(methacrylic acid) in aqueous solution, he suggested that this hydrophobic interaction stabilised the swelling of PHEMA hydrogel in water (always about 40% water) independent of its history. He confirmed this hypothesis by adding microsolutes to the equilibrating solutions of PHEMA hydrogel and observing a swelling or deswelling of the hydrogel, depending on the type of the solute added, which he interpreted as being due to breaking or enhancing respectively the hydrophobic interactions in the polymer by the solute. Refojo even went further to make use of this hypothesis to explain the transient opacity observed during exchange of a good solvent for water in PHEMA gel.

Later, Gouda et al[120] provided evidence for this hypothesis of alternating hydrophilic and hydrophobic layers in PHEMA hydrogel, which they called a micro-mesomorphic order of either nematic or smectic type. They found that although PHEMA hydrogel was macroscopically transparent, it exhibited a strongly angle dependent anisotropic scattering which they explained must be due to existence of an ordered structure probably induced by the hydrophobic forces in the polymer. However even in pure ethylene glycol which is a good solvent for PHEMA, the PHEMA gel was found to show an angle dependent anisotropic scattering and they interpreted this remaining anisotropic scattering as being due to inhomogeneous cross-linking.

Further evidence was provided by Ilavsky et al[121] on the rheo-optical (birefringence relaxation) studies of PHEMA gels. They concluded that some structural order was induced in the gel by the diluent, in addition to inhomogeneous cross-linking, and this concept exceeded that implied by a Gaussian network which postulates randomly

coiled polymer chains in a randomly cross-linked network. However they believed that such layered regions were limited in size and probably dependent on the crosslink density, and that the presence of cross-links and the ataticity of the chains most likely impeded the development of crystallinity.

I.5.B. Polymerisation and Copolymerisation

Having a vinyl group in its structure, the monomer HEMA can be polymerised under suitable conditions, just like methyl methacrylate. The preparation of the three-dimensional PHEMA from the monomer is carried out by copolymerisation of HEMA with its diester EGDMA (i.e. ethylene glycol dimethacrylate) using a free radical initiator (e.g. azo compounds, peroxides and redox catalysts). Although thermal polymerisation is normally used, photopolymerisation of HEMA has also been reported[122]. In the contact lens application, the concentration of the cross-linking agent used is normally less than 1% of the weight of the monomer so that the polymer obtained is sparingly cross-linked, and the resulting hydrogel is elastic (or "springy"). The polymerisation of HEMA with EGDMA can be achieved in bulk or in water or a good solvent (e.g. ethylene glycol). It has been found that in aqueous medium[115,121], depending on the concentration of water in the starting mixture, the polymerisation process yields either a homogeneous (< about 40% water in the starting mixture) or a heterogeneous hydrogel (> about 50% water). The formation of a heterogeneous structure was explained by phase separation of the polymer from water owing to water being a thermodynamically poor solvent for PHEMA. Polymerisation in thermodynamically better solvents (e.g. ethylene glycol or ethanol) leads eventually to a homogeneous hydrogel. Increasing the amount of cross-

linking agent (EDGMA) may also result in a heterogeneous polymer, but this influence is much less important than that of solvent.

Copolymerisation of 2-hydroxyethyl methacrylate (or its related compound) with other monomers have been reported but there are not many data available in the literature on the values of the copolymerisation parameters (such as reactivity ratios, Q and e values). However the data published so far are shown in Tables I.3 and I.4.

Table I.3.

Reactivity Ratios.

Monomer M_1	Monomer M_2	Reactivity ratio r_1	Reactivity ratio r_2	Temp (°C)	Ref.
hydroxyethyl methacrylate	allyl chloride	35.6	0.033	60	124
ethylene glycol dimethacrylate	styrene	0.65	0.35	60	125
ethylene glycol dimethacrylate	acrylonitrile	1.25	0.22	60	126

Table I.4.

Q and e Values

Monomer	Q	e	Ref.
hydroxyethyl methacrylate	0.80	0.20	127
hydroxypropyl methacrylate	0.79	0.20	
ethylene glycol dimethacrylate	0.88	0.24	

I.5.C. Swelling

The presence of the hydroxyl groups in PHEMA gives rise to the uptake of water when they are in contact with each other, and the existence of the three-dimensional network of the polymer restricts the polymer from overhydration or dissolution in water. There exists a point when the tendency to swell, caused by the osmotic pressure of the polymer segments, is equally opposed by elastic retractive forces arising as the chains between cross-links elongate. The water uptake by the polymer at this equilibrium point is called the equilibrium water content of the polymer.

The equilibrium water content of transparent PHEMA hydrogel has been found to be around 40% at 25°C, depending on the experimental conditions, cross-link density, etc. This low value was explained by the thermodynamically poor solvent power of water for the polymer. Swelling of PHEMA increases in polar liquids which are better solvents than water (such as dimethylformamide and ethylene glycol) or in a mixture of water and a thermodynamically better solvent, (e.g. water/ethylene glycol mixture).

Wichterle et al[129] have studied the volume swell of PHEMA gels equilibrated in various solvents, and their results are shown in Table I.5. It can be seen from the table that dimethylformamide is the best swelling agent for cross-linked PHEMA, and has a swelling power of about 3½ times that of water.

While PHEMA takes up about 40% water on hydration, the equilibrium water content of transparent poly(hydroxypropyl acrylate) hydrogel was found to be around 50%[128] and that of poly(glycerol methacrylate) hydrogel around 75%[130].

There are several factors which have been found to affect the equilibrium water content of a hydrogel.

Table I.5.

Volume swelling of PHEMA (0.31% of EGDMA) in various solvents[129]

Solvent	Volume of gel / Volume of polymer	Solvent	Volume of gel / Volume of polymer
methyl alcohol	3.04	ethylene glycol monoethyl ether	4.95
ethyl alcohol	2.22	ethylene glycol monoethyl ether	3.74
propyl alcohol	1.99	ethylene glycol monobutyl ether	1.18
butyl alcohol	1.07		
tert-butyl alcohol	4.36	glyceryl isopropylidene ether	1.12
allyl alcohol	3.79	ethyl acetate	1.14
tetrahydrofurfuryl alcohol	4.91	ethylene glycol dimethacrylate	1.21
cyclohexyl alcohol	1.18	2-hydroxyethyl methacrylate	3.04
ethylene glycol	3.80	water	1.68
diethylene glycol	4.52	toluene	1.16
triethylene glycol	4.18	chloroform	1.29
glycerol	1.14	formic acid (90%)	5.18
acetone	1.58	formamide	4.03
methyl ethyl ketone	1.37	dimethyl formamide	6.15
cyclohexanone	1.20	pyridine	5.50
dioxane	1.16	dimethyl sulphoxide	5.85
tetrahydrofuran	1.65		

I.5.C.1. Effect of Initial Dilution and Solvent Composition

As the monomer HEMA is miscible with water to a certain extent, polymerisation of HEMA has been carried out in aqueous medium, and it has been found[123,131] that when up to about 40% water was used in the polymerisation mixture, the PHEMA hydrogel obtained was transparent and its final equilibrium water content was about 40%, and that if water used in the polymerisation mixture was more than 40%, the polymer started to precipitate during polymerisation, and a translucent or an opaque hydrogel (or sponges) was obtained, depending on the concentration of water in the polymerisation mixture.

Refojo et al[128] found that ethylene glycol was a better solvent than water for PHEMA, and by using a mixture of ethylene glycol and water as the diluent in the polymerisation of HEMA, and then equilibrating the gel obtained in water, they could increase the diluent concentration in the polymerisation mixture up to at least 63% without obtaining a heterogeneous gel during polymerisation (compared with 40% if water alone was used as the diluent), but the final hydrogel had still an equilibrium water content of about 40%. Further investigation by Yasuda et al[131] showed that the equilibrium water content of the PHEMA hydrogel did not depend on the composition of the diluent in which polymerisation took place, but the critical initial dilution, at which precipitation of the polymer occurred, increased with the increasing amount of ethylene glycol in the ethylene glycol - water diluent mixture. They observed that those gels, when equilibrated in water after completion of the polymerisation, swelled or deswelled to reach their final equilibrium water contents accompanied by a volume change, but with use of approximately 35% solvent in the polymerisation mixture of HEMA, regardless of the composition of the

solvent, an "isovolumic" gel could be obtained which did not change its volume when equilibrated in water after completion of gel formation.

The equilibrium water content of transparent PHEMA hydrogel was found to be less than 41%, translucent between 41% and 54% (the turbidity increased with the water content), and opaque higher than 54%[131].

Contrary to the independence of equilibrium water content of homogeneous PHEMA hydrogel on the initial dilution, Refojo[130] found that the equilibrium water content of transparent poly(glycerol methacrylate) (i.e. PGMA) hydrogel increased linearly with the amount of water used in the polymerisation mixture, and attributed this to the solubility of poly(glycerol methacrylate) in water. He concluded that the degree of swelling of the polymer at the time of network formation affected the final equilibrium water content of the hydrogel. However, this does not apply to PHEMA hydrogel.

Spin-cast PHEMA contact lenses have been produced in the presence of a good solvent such as ethylene glycol. On equilibrating in saline solution, the lens contracts, and it has been claimed[29] that by using this technique, surface irregularities on the contact lens can be reduced and a better lens is thus obtained.

I.5.C.2. Effect of Cross-linking

On investigating the effect of the amount of cross-linking agent upon equilibrium water content, Refojo et al[128] found that the equilibrium water content of PHEMA hydrogel remained quite constant (about 40%) over a range of 3% cross-linking agent, triethylene glycol dimethacrylate (TEGDMA). They also found that the equilibrium water content of poly(hydroxypropyl acrylate) (i.e. PHPA) hydrogel decreased

by 4% as the concentration of TEGDMA increased to 3%, and suggested that TEGDMA was more soluble in the HPA - ethylene glycol - water system than in the HEMA system, therefore making it possible to introduce more cross-links in the HPA polymer. It was also found[130] that the equilibrium water content of PGMA hydrogel decreased from 75% to 61% if the TEGDMA in the monomer mixture was increased from 2% to 10%.

In studying the effect of cross-linking on the equilibrium water content of PHEMA hydrogel equilibrated in 3% saline solution and artificial tears, Yasuda et al[131] observed that the reduction was not great and the effect seemed to level off rapidly with the increasing amount of cross-linking agent.

I.5.C.3. Effect of Initial Concentration.

The equilibrium water content in a hydrogel has been found to be affected by initiator concentration[131]. For PHEMA hydrogel, the equilibrium water content increased with increasing initiator concentration and was explained by relating to the number of growing radicals.

I.5.C.4. Effect of Hydration Temperature

On investigating the effect of hydration temperature on the equilibrium water content of PHEMA hydrogel, Refojo et al[128] observed a minimum in the water content (around 37%) equilibrated at about 60°C, and the level of this minimum water content was affected by the temperature at which polymerisation of HEMA was carried out. They also observed that the equilibrium water content of PHPA hydrogel decreased steadily with increasing temperature up to 30°C, above which the equilibrium water content approached a constant value of 33%. Swelling

of PGMA hydrogel was found[130] to behave in the opposite way: it swelled increasingly with temperature. It was suggested that swelling of PGMA hydrogel was an endothermic process, and swelling of PHPA hydrogel was an exothermic process.

I.5.C.5. Effect of Equilibrating Solution

The presence of a solute in water, in which a hydrogel is equilibrated, has been found to affect the equilibrium water content of the hydrogel. Yasuda et al[131] obtained a lower equilibrium water content of PHEMA hydrogel with increasing concentration of saline (NaCl) solution equilibrating the hydrogel. On the other hand, an increase in water content has been observed[128] when the hydrogel is kept in an aqueous fluorescein solution. Also Refojo[119] observed a swelling of PHEMA hydrogel in the presence of urea and its methyl derivatives, organic solvents like alcohol and acetone, and anions like iodide, acetate, trichloroacetate and thiocyanate, but on the other hand, a deswelling in the presence of chlorides, sulphates, bromides and cetyl pyridinium chloride.

In contact lens applications, a hydrogel contact lens will maintain its shape, size and refractive index only when the variables affecting its degree of swelling remain constant. The variables include ambient temperature and relative humidity, rate of blinking and tearing, and nature and concentration of solutions bathing the lens.

I.5.D. Optical Properties

It has been found[132] that the temperature dependence of refractive index (n) for a dehydrated poly(2-hydroxyethyl methacrylate) polymer at temperature t is given by $n = 1.51250 - 0.000156\ (t - 20)$, and that the refractive index of a swollen PHEMA gel depends on the nature of the swelling agent and on the degree of swelling. Refojo

et al[128] obtained a value of 1.431 for PHEMA hydrogel with an equilibrium water content of 40%.

When HEMA is polymerised in the presence of a supercritical amount of a poor solvent (e.g. > 40%water), phase separation occurs during polymerisation and the hydrogel becomes translucent or opaque, and this was explained that regions with different refractive indices were formed. In studying the effect of temperature on the equilibrium water content of PHPA hydrogel, Refojo et al[128] observed that the hydrogel became opaque at temperatures above 30°C if the change in temperature was rapid, and regained its transparency if allowed to stand long enough at the higher temperature. They interpreted the opacification of PHPA hydrogel upon rapid changes in temperature as a microphase separation due to uneven configuration changes of polymer segments caused by temperature changes which were faster than the relaxation time for the configurational change of polymer segments. Also cooling of a 1-butanol swollen PHEMA gel at a high temperature has been found to induce heterogenity[132]. The exchange of a good solvent (i.e. ethylene glycol) in PHEMA gel for water has also been found[128,133] to cause transient opacity, but the transparency was regained after equilibrium was reached. This phenomenon has been called "microsyneresis", and Refojo[119] interpreted this as being caused by the inability of water to solvate the hydrophilic ends of the unorganised polymer segments, but homogeneity returned to the hydrogel after a rearrangement of the chains, directed by the interaction of the hydrophobic portions of the polymer segments, exposing to water most of the hydrophilic sites in the network. Despite transparent macroscopically, PHEMA hydrogel has been found to be microscopically birefringent[120,121]. However the transparecny of PHEMA hydrogel enables it to be used in the contact lens field besides biomedical applications.

I.5.E. Permeation and Diffusion

The rate of transport of low molecular weight compounds (e.g. water, oxygen, sodium chloride) through hydrogels is an important property in many applications in biological systems.

The permeation of oxygen through hydrogels is particularly important in contact lens application. Unfortunately the oxygen permeability coefficients of PHEMA hydrogel which have been reported in the literature by many workers are not consistent with each other but vary by at least 13 times, and so the actual value has been the centre of dispute for many years. The first value of the oxygen permeability coefficient of PHEMA hydrogel since its introduction as a contact lens material was reported by Takahashi and his co-workers[87] as not more than 4×10^{-10} cc.(S.T.P.). mm.cm^{-2}.sec^{-1}. cm.Hg^{-1}. Later Yasuda and his collaborators[131,134] determined the oxygen permeability coefficient of PHEMA hydrogel as 51×10^{-10} cc (S.T.P.). mm. cm^{-2}.sec^{-1}. cm Hg^{-1}. Kamath[135] in his paper on ideal contact lenses, reported a value of 4×10^{-10} cc (S.T.P.). mm. cm^{-2}.sec^{-1}. cm Hg^{-1}, so did the paper written by Fatt et al[61]. However Fatt and his co-workers[136] recently determined a value of 89×10^{-10} cc (S.T.P.). mm.cm^{-2}.sec^{-1}.cm Hg^{-1} at 25°C for Soflens contact lens (i.e. PHEMA).

The diffusion coefficient of oxygen in PHEMA hydrogel has been determined by some Czech workers[86], who found that the oxygen diffusion coefficient was very sensitive to changes in the hydrogel structure, particularly the degree of crosslinking. The diffusion coefficient was found to decrease with increasing concentration of cross-linking agent used. In the investigation of hydroxyethyl methacrylate and glycerol methacrylate (i.e. GMA) copolymers, Yasuda et al[131] found an increase in the oxygen permeability coefficient of

the copolymer hydrogel as the glycerol methacrylate content in the polymer was increased. They also observed an abrupt increase in oxygen permeability when the PHEMA hydrogel became translucent or opaque.

The permeability of PHEMA hydrogel to water has been studied and compared with other materials like PHPA hydrogel, PGMA hydrogel, cellophane and polyelectrolyte complexes. From the measurements of water permeation, Refojo[137,138,139] calculated the diffusion coefficient of water in the hydrogels and their "pore radii", and his results, as shown in Table I.6, indicate a water diffusion coefficient of 3.08×10^{-5} $cm^2.sec^{-1}$ and a "pore radius" of about 0.4mμ for PHEMA hydrogel with an equilibrium water content of 38.7%. It was thus concluded that the flow of water in the PHEMA hydrogel was diffusive and similarly the PHPA hydrogels (40 to 60% water), but the water movement in PGMA hydrogels (74 to 94% water) and polyelectrolyte complex hydrogels were predominantly viscous flow.

TABLE I.6.

Water permeability coefficients, diffusion coefficients and "pore radii" of some hydrogels[137]

Hydrogel	Equilibrium water content (%)	Water permeability coefficient $Kx10^{15}$, cm^2	Water diffusion coefficient $Dx10^5$, $cm^2/sec.$	Pore radius mμ
PHEMA	38.7	0.075	3.08	0.40
PHPA	41.2	0.087	3.89	0.45
	62.6	0.28	7.76	0.63
PGMA	80.3	4.65	88.6	2.17
	83.2	6.08	132.8	2.62
	88.8	14.79	265.0	3.70
	93.6	42.04	750.3	6.23
Cellophane	57.6	1.044	-	-
Polyelectrolyte complex	66.6	8.18	255.7	3.64

The idea of "pore radius" of a hydrogel has been suggested to provide a rough indication of the size of molecules that can and cannot penetrate the network. Although a pore radius of about 4 Å has been obtained for homogeneous PHEMA hydrogel, Refojo[140] reckoned that this value might be too small because fluorescein of molecular radius 5.5 Å could penetrate into the PHEMA hydrogel, the result of which fluorescein cannot be used during fitting of hydrogel contact lenses. However increasing the molecular size of a solute reduces the ease of penetration.

The permeation of water through PHEMA hydrogel can be increased by incorporation of silica fillers, as investigated by Refojo[141]. He explained that both the amount of water at equilibrium and the amount of filler in the hydrogel contributed to the increase in water permeability of the PHEMA membranes. Because of high permeability of water and permselectivity, these heterogeneous PHEMA membranes have been suggested for desalination of water by reverse osmosis. Alternately porous PHEMA hydrogels can be made by a special polymerisation technique suggested by Haldon and Lee recently[142]. They cooled the monomer mixture so that the solvent froze and the monomer occupied the interstices between the solvent crystals, before photopolymerisation was started. The porous PHEMA films thus obtained have different pore radii and hence different water permeabilities (but higher than the normal PHEMA hydrogel), depending on the experimental conditions such as the rate of freezing, the profile of thermal gradient, and the solvent and cross-linked concentrations.

Earlier, White[143] has shown that the water permeability of cross-linked acrylamide hydrogel with a relatively low cross-link density depends on the amount of polymer in the hydrogel and is

independent of the amount of cross-links in the network. In studying the water permeation in other homogeneous hydrogel membranes such as poly(glycerol methacrylate) hydrogels, Yasuda et al[144,145] found that at small hydration (volume fraction of water < 0.2), water permeated by diffusion even under an applied hydraulic pressure gradient, but in the higher hydration region, water permeated by bulk flow under an applied hydraulic pressure gradient but moved only by diffusion in the absence of a pressure gradient. But very recently Paul[146] argued that the equation used by the previous workers to compute diffusion coefficients from hydraulic permeabilities for a homogeneous water-swollen membrane might be in error. He developed a substitute equation and concluded that molecular diffusion was predominant in homogeneous hydrogels even at high hydration.

However the permeability of hydrogels to water is an important property in regard to their potential use as implants in surgery. In fact some high water content, transparent hydrogels have been suggested as potential replacement of the cornea or part of the cornea[138].

The permeation of hydrogels to electrolytes such as sodium chloride is important where desalination and haemodialysis are involved. The permeability of PHEMA hydrogel and other hydrogels to sodium chloride has been investigated by Yasuda and his collaborators[147,148], who found that the NaCl permeability increased exponentially at higher range of hydration, but at lower range of hydration the permeability value was affected by the distribution coefficient of sodium chloride in the membrane and solution. Investigation into the effect of chemical structure of hydrogel polymer on the NaCl permeability by Kopecek et al[1] showed that introduction to an ionogenic group (e.g. methacrylic acid or 2-(diethyl-amino) ethyl methacrylate) into PHEMA hydrogel decreased

the NaCl permeability whereas an ampholytic hydrogel had a higher NaCl permeability than a neutral one, which they explained as being due to an increase in the diffusivity of the salt in the ampholytic structure. Recently Kopecek et al[150] found that replacement of the HEMA in the HEMA - MMA copolymer by a more hydrophilic monomer 2-(hydroxyethoxy) ethyl methacrylate increased the permeability of the hydrogel to sodium chloride at the same degree of hydration.

The diffusion and permeation of PHEMA hydrogel to other solutes such as potassium chloride[151], magnesium sulphate[149] and organic low molecular weight compounds such as urea[148] ethanol and cyclohexanol[152] have also been studied. Some of the properties are particularly important in haemodialysis application of hydrogels.

Molecules such as glucose and amino-acids can penetrate PHEMA hydrogel, but large molecules such as polysaccharides cannot. The fact that PHEMA hydrogel allows penetration of small molecules but not large molecules leads to its potential use in gel filtration[153].

When a PHEMA hydrogel is immersed in a solution of a solute, it absorbs the solute and the concentration of which in the gel matrix will depend on the affinity of the solute to the polymer matrix. Absorption of some substances (e.g. preservatives used in the contact lens soaking solutions) by hydrogel contact lenses and subsequent release into the wearer's eye may be harmful, if the substances are toxic to the eye tissues. But on the other hand, these properties of absorption and desorption of PHEMA hydrogel have been utilised to achieve a slow release of physiologically active substances incorporated in the gel[107-109, 154-156]. The incorporation occurs by diffusion from solution to the gel, by incorporating the substance into the monomer mixture, or by drying a solution containing the

substance and an uncross-linked polymer[107].

I.5.F. Chemical Properties

PHEMA hydrogel is known to be chemically stable. Model compounds, low molecular weight ester of ethylene glycol and diethylene glycol with pivalic acid, were found[157] to be very resistant to hydrolysis. Alkaline hydrolysis in sodium hydroxide solution[158] of diethylene glycol methacrylate has been investigated and found to be much faster than acid hydrolysis in sulphuric acid[159].

Acetylation of soluble PHEMA[160] and other reactions of cross-linked PHEMA with amines, hydrazine, hydroxylamine, and acetic anhydride[159] have also been investigated.

I.5.G. Mechanical Properties

In the dry state, poly(2-hydroxyethyl methacrylate) is hard and brittle, resembling poly(methyl methacrylate). As a result, PHEMA can be lathe-cut successfully in contact lens productions. When hydrated, PHEMA hydrogel is soft and rubbery, and this property is important when the hydrogel is used in contact with soft tissues, such as the cornea. However in the hydrated state, PHEMA hydrogel exhibits low tensile properties and low tear strength. Similar properties have been reported for hydrogels of polyelectrolyte complexes.

Apparently synthetic hydrogels such as glycol methacrylate hydrogels are a different class of polymeric material from most of the conventional plastics and rubbers. The mechanical properties such as creep, stress relaxation, equilibrium deformation behaviour and ultimate properties of glycol methacrylate hydrogels (particularly PHEMA hydrogel) have been extensively studied, particularly in relation

to chemical structures, by Czechoslovak workers. However some of the mechanical properties of PHEMA hydrogel and its related compounds are described as follows.

The creep properties of dry cross-linked PHEMA have been studied in the main transition region and compared with those of poly(propyl methacrylate) by Ilavsky et al[161], who concluded that the softening caused by the side chain in PHEMA was outweighed by increased intermolecular forces of hydroxyl groups. Copolymers of HEMA and other monomers such as diethylene glycol methacrylate[162] and butyl methacrylate[163] at various compositions have also been studied, and the creep properties of these copolymers were found to shift towards shorter retardation times than PHEMA with increasing comonomer content, and were interpreted as being due to an increase in the free volume and hence the mobility of the chains in the copolymers. The viscoelastic properties of dry PHEMA in the main transition region have been compared with other esters of poly(methacrylic acid), and it was found[164] that the replacement of the hydroxyl group in the side chain by a methyl group resulted in a decrease of the monomer frictional coefficient and the glass transition temperature (T_g) whereas replacement of the ethereal oxygen in the side chain by a methylene group has no effect on the monomer frictional coefficient and the glass transition temperature, which are shown in Table I.7.

TABLE I.7.

Glass transition temperatures and densities of some methacrylates[164]

Structure	Polymer	T_g (°C)	Density (g/cm^3)
$CH_2 = C(CH_3)COO(CH_2)_nH$ (where n = 3,4,6)	(poly(propyl methacrylate)	43.0	1.083
	(poly(butyl methacrylate)	19.0	1.055
	(poly(hexyl methacrylate)	-17.0	1.013
$CH_2=C(Me)COO(CH_2CH_2O)_nH$ (where n = 1,2)	(poly(2-hydroxyethyl methacrylate)	93.5	1.286
	(poly(diethylene glycol methacrylate)	11.0	1.256
$CH_2=C(Me)COOCH_2CH_2O(CH_2)_nH$ (where n = 1,3)	(poly(methoxyethyl methacrylate)	20.0	1.174
	(poly(propoxyethyl methacrylate)	-20.0	1.088

The glass transition temperature (T_g) of dry PHEMA has also been determined dilatometrically to be 98°C[162] or 86°C[165,166].

Swelling agents such as diethylene glycol have been found to shift the viscoelastic properties of PHEMA towards shorter retardation times in the main transition region, both by creep measurements in the low diluent contents[167] and by dynamic measurements in the high diluent contents[168], and this was explained in terms of an increase in free volume. Investigations have also been made in the rubberlike region of the creep of the PHEMA in the dry state[169] and in the diethylene glycol swollen state[170,171, 172].

The relaxation behaviour of PHEMA (dehydrated) below the glass transition temperature has been studied and found to have a low temperature loss maximum γ at the temperature T_γ = -125°C (at 1 Hz)[173] and also a secondary loss maximum β at the temperature T_β = 30°C (at

1 Hz)[174,175], attributed to the relaxations of the $-CH_2CH_2OH$ group and -COOR group respectively. The T_γ was found[176] to be unchanged as the length of the polar side chain was increased. However an increase in the concentration of polar low molecular weight compounds in PHEMA was found to shift the main transition region towards lower temperatures[167,168], accompanied by a gradual disappearing of the γ dispersion[174,177] and a gradual appearing of a new loss maximum $\beta_{(sw)}$ at -70°C (at 1 Hz)[178], which was shifted towards lower temperatures (with increasing intensity) with increasing degree of swelling. This was interpreted as a consequence of the relaxation motions of side chains interacting with the molecules of the low molecular weight compound such as water or diethylene glycol.

The relaxation behaviours of heterogeneous PHEMA hydrogels below the glass transition temperature have also been studied[179] and it was concluded that the heterogeneous hydrogel at low temperatures could be regarded as a biphasic system in which one component was represented by the polymer with a bound "monolayer" of water and the other was ice.

The thermomechanical properties of dry PHEMA have been studied[180], in relation to the polarity and length of the side chain, and to the cross-link density. A relation was obtained between the physical degree of cross-linking and the content of cross-linking agent added.

From the study of equilibrium deformation (tensile) behaviours of PHEMA, it has been established[181,182,183] that increasing concentration of a diluent during polymerisation decreased the cross-linking efficiency. The kinetic theory of elasticity for rubber was applied to study the equilibrium deformation behaviour, and deviation from the theory on PHEMA gels was recently interpreted as being due to inhomogeneous cross-linking and diluent-induced structure order in the

gels[121,171].

The ultimate tensile behaviour of PHEMA hydrogel at equilibrium hydration has been studied[184], and found to be sensitive to temperature and rate of elongation. The degree of dilution during polymerisation also affects the ultimate properties of PHEMA : the increasing concentration of the diluent results in a decrease in the ultimate properties and a slight increase in elongation at rupture, but when heterogeneous gels are obtained, the tension strength decreases rapidly. The variation of the ultimate behaviours of PHEMA gels at various temperatures has also been investigated[185,186].

A comparison between the ultimate properties of PHEMA hydrogel and polyelectrolyte complex hydrogel has been made by Bixler et al[187], and their results are shown in Table I.8.

Table I.8.

Some mechanical properties of three types of hydrogels[187]

Hydrogel	Tensile strength (psi)	Modulus (psi)	Elongation (%)
PHEMA (45% water)	60	1,000	140
polyelectrolyte complex (55% water)	800	8,000	18
cellophane (55% water)	4,100	4,400	92

I.5.H. Soluble Polymer of Hydroxyethyl Methacrylate

Soluble uncross-linked PHEMA has been prepared at low conversion of polymerisation or synthesised in a thermodynamically good solvent (such as dimethylformamide, dimethyl sulphoxide, methanol or ethylene glycol) which acts as a diluent to reduce chance of cross-linking[115,188].

The molecular weight of soluble PHEMA has been determined[189] from the intrinsic viscosity measurements in dimethylformamide and found to satisfy the equation $[\eta] = 8.90 \times 10^{-5} . M^{0.72}$ (in dl./g.) at 25°C in the molecular weight range $5 \times 10^4 < M < 7 \times 10^6$.

Uncross-linked PHEMA is soluble in polar solvents such as methanol, dimethylformamide and dimethyl sulphoxide[190]. The uncross-linked PHEMA is not soluble in water but soluble in a cosolvent composed of two non-solvents (e.g. dioxane and 10 - 78% water mixture)[191], and also in aqueous solutions of some inorganic salts (e.g. perchlorates of Na, Mg, iodides of Na, K).

Uncross-linked HEMA polymers have been used in the preparation of surgical sutures[192], antifogging coatings[193], and hydrophilic varnishes and paints[193].

I.6. Desired Characteristics of a Continuous-wear Contact Lens Material.

It has been suggested that a material for biomedical applications should have the following characteristics:

(1) it can be reproducibly obtained;

(2) it can be fabricated into any shape without being degraded or adversely changed;

(3) it should have the needed chemical, physical and mechanical properties to perform its function;

(4) it can be sterilised (heat or chemical) without change in properties or dimensions;

(5) it must not have its properties adversely altered by the biological environment; and

(6) it must be toxicologically inactive in its biological environment.

Likewise all these characteristics are also required by a continuous-wear contact lens material. An ideal contact lens has been defined by Kamath[135] as one possessing five functional criteria of comfort and safety, acuity, wearing ease, stability and care. He further compared the physical and chemical properties of PMMA, PHEMA and silicone rubber contact lenses with those of the cornea, and worked out the physical and chemical attributes of his ideal contact lens to satisfy the criteria he set. His results are shown in Table I.9. Kamath's idea of designing an ideal contact lens material by using the cornea as a model is interesting and reasonable. However he did not stress the importance of the requirement of very high oxygen permeability in an ideal contact lens for continuous wear.

Because of good tolerance with soft tissue, hydrogels seem to be the right class of material for contact lens applications. PHEMA hydrogel satisfies some of the requirements but apparently does not permeate sufficient oxygen for the continuous wear purpose, as it has been used for daily wear only. Mandell and Polse[194] found that wearing contact lenses during sleep showed considerable corneal swelling and undesirable tissue changes and attributed this to the reduction in the oxygen available to the corneal epithelium from the atmosphere. Recently continuous wear of PHEMA contact lens (Soflens) has been attempted[195], but edema and corneal thickening were observed in those patients wearing Soflens day and night for 10 days.

By studying the steady-state distributions of oxygen and carbon dioxide in the in-vivo cornea of an eye covered by a tight gas-permeable contact lens, Fatt et al[61] concluded that (i) silicone rubber lenses up to 0.2 mm thick, even when tightly fitted, would allow normal oxygen tensions in the cornea and could be worn even during sleep, and

Table I.9.

Kamath's ideal contact lens[135]

Property	Cornea	PMMA	PHEMA	Silicone rubber	Kamath's ideal contact lens
(1) Density	1.03	1.18	1.1	1.16	~ 1
(2) Modulus of rigidity, (dynes/cm^2)x10^{-7}	13	3,100	0.2	4	> 150
(3) Tensile strength (dynes/cm^2)x10^{-7}	7	60	0.3	11	> 50
(4) Tear strength (gms/mm)	2,000	brittle	10	2,000	~ 1,000
(5) Optical transmission (%. cm^{-1})	> 98	96	>95	< 95	> 96
(6) Refractive index	1.37	1.49	1.4	1.43	~ 1.4
(7) Surface hardness	Shore A.15	Rockwell M.90	Shore A.19	Shore A.96	> Rockwell M.90
(8) Surface finish (Å)	50	500	500	-	300
(9) Surface character	hydrophilic	hydro-phobic	hydro-philic	hydrophobic	Hydrophilic
(10) Oxygen permeability (cm^4/sec.dyne) x10^{15}	400	0.7	3	1,000	> 60
(cc.mm/cm^2.sec. cm Hg) x 10^{10} *	520	1	4	1,300	> 80
(11) Water permeability cm^4/sec.dyne)x10^{15}	100	0	10	0	preferably low
(12) Chemical nature	collagen protein	ester polymer	HEMA polymer	dimethyl siloxy polymer	any inert high polymer
(13) Water content (%)	81	0.4	45	0.12	< 2

* Computed from Kamath's values by using

$$1\ cm^4.sec^{-1}.dyne^{-1} = 1.3x10^5\ cc.mm.cm^{-2}.sec^{-1}.\ cm\ Hg^{-1}.$$

(ii) ethyl cellulose and the more permeable polystyrene would also probably be useful as tight contact lens if the thickness was kept at 0.1 mm. But tight contact lenses are not encouraged by most opticians because an existence of the tear fluid between a contact lens and the cornea will not only help the access of oxygen to the corneal epithelium but also act as a reservoir for the metabolic products of the cornea (e.g. lactates, cell debris) which will then be discharged during blinking.

However it is reckoned that a combination of the two approaches (i.e. using a contact lens material of high oxygen permeability and allowing an air-saturated tear pool between the contact lens and the cornea) will provide a greater advantage in designing an ideal contact lens for continuous wear. While the second approach requires a good fitting of a contact lens and is an optician's job, the first approach involves the chemistry of the contact lens material and is particularly important during sleep when there is very little movement of the eye and blinking of the eye-lid, and when the oxygen tension at the back of the lens is only 55mm Hg which is about one-third of that in an open eye.

I.7. Scope and Object of the Present Work

It is considered that covalently cross-linked synthetic hydrogels are in many ways very suitable for contact lens materials, as mentioned in the previous sections. Poly(2-hydroxyethyl methacrylate) hydrogel was originated from an idea of synthesising polymeric materials for biomedical applications[4], and not purposedly designed as a contact lens material. One of the drawbacks of this material in contact lens applications is its insufficient oxygen permeability to meet the

metabolic requirements of the cornea. As a continuous-wear contact lens is defined as one which can be worn in the eye continuously during successive day and night periods without causing any adverse effect on the eye, a contact lens material for continuous wear must possess an oxygen permeability sufficiently high to meet the oxygen consumption rate of the cornea. How high the oxygen permeability of a continuous-wear contact lens must be required has not been well-defined although Fatt et al[61] suggested a minimum oxygen permeability coefficient of about 250×10^{-10} cc(S.T.P.).mm.cm^{-2}.sec^{-1}. cm Hg^{-1} for a tight contact lens. However it is reckoned[196] that a material for a continuous-wear contact lens must have an oxygen permeability coefficient of at least 200×10^{-10} cc(S.T.P.).mm.cm^{-2}.sec^{-1}.cm Hg^{-1}, although this value has not been confirmed clinically. Reviewing those commercially available plastics and rubbers, not many polymers have such a high oxygen permeability coefficient as well as required optical properties (e.g. silicone rubber, other rubbery materials), but they are expected to be incompatible with the cornea because of their hydrophobic nature.

To "tailor-make" a polymer which can be used as a contact lens material for continuous wear, bearing particularly in mind the oxygen requirement of the cornea for its metabolism, necessitates first the correlation of chemical structures and physical properties of polymers. One of the quicker and perhaps more profitable approaches to the problem is to synthesise copolymers which provide a balance of the properties of the corresponding homopolymers.

As it might have been realised, it is difficult to define absolutely hydrogels and non-hydrogels in terms of equilibrium water content. However it is reckoned here that a hydrogel is one which contains more than 10% water at equilibrium hydration, although this

value is arbitrary. As water is one of the constituents in a hydrogel, it is expected that water may also play a part in controlling the properties of the hydrogel. Again, structural and composition variations by copolymerisation may affect the equilibrium water content of a hydrogel.

To provide hydrophilicity in an ideal contact lens material, at least one of the chemical constituents in the copolymer must be hydrophilic. So copolymerisation of a hydrophilic monomer and a hydrophobic monomer, and of two hydrophilic monomers with structural variations should give a wide range of copolymers.

The hydrophilic monomers used in the present work include glycol methacrylates (such as hydroxyethyl methacrylate, hydroxypropyl methacrylate), glycol acrylates (such as hydroxyethyl acrylate, hydroxypropyl acrylate) and N-vinyl pyrrolidone, while the hydrophobic monomers used include styrene, para-methyl styrene, 4-methyl pent-1-ene and some alkyl methacrylates such as methyl methacrylate.

Hydroxyethyl methacrylate is chosen as a suitable starting monomer in copolymerisations because the chemistry, properties and non-toxicity of PHEMA hydrogel are quite known. Although the oxygen permeability value of PHEMA hydrogel has not been well-defined and varies by at least 13 times between various workers[87,134], it is hoped that the present work may help to contribute to the solution of the argument. Other hydroxyl containing methacrylate and acrylates are included in the present study to provide structural variations. Poly(vinylpyrrolidone) has been used as a blood plasma[197] and so vinyl pyrrolidone copolymers are expected to be biocompatible.

Styrene is included in the present work as a hydrophobic monomer for several reasons: (1) a high oxygen permeability coefficient of 250×10^{-10} cc (S.T.P.). mm.cm^{-2}.sec^{-1}. cm Hg^{-1} has been reported for

polystyrene by Rogers[198] and Fatt et al[61] who in fact suggested that polystyrene might be a suitable material for tight contact lenses, (2) polystyrene is being used in food packaging industry, suggesting that the polymer is not toxic, and (3) styrene has been commercially and abundantly available for many years and is known to be easily polymerisable. Poly(4-methyl pent-1-ene) has been known as a transparent plastic material for some years and reported to have quite a high oxygen permeability coefficient, and so the monomer 4-methyl pent-1-ene is included in the copolymerisation study. On the other hand, although poly(methyl methacrylate) is reported to be very impermeable to oxygen, the monomers such as methyl methacrylate and other methacrylates are included here to provide structural variations. However copolymerisation will be further extended to terpolymerisation if necessary in order to achieve certain properties.

Synthesis of copolymers (and terpolymers) will be carried out in bulk and, in some cases, in solution. In bulk polymerisation the end product is in a sheet form which enables physical properties to be measured, or in a rod form which can be subsequently lathe-cut into contact lenses. In solution polymerisation, the end product will be in a powder form, and the possibility of compression moulding contact lenses from the polymer powder will be preliminarily investigated.

With the hydrogels eventually obtained in sheet forms, physical properties will be evaluated so that structures and properties can be correlated. Since it is unlikely to study all the physical properties of hydrogels in the present work, some properties which are thought to be important in contact lens applications are selected, such as equilibrium water content, density, refractive index, oxygen permeability and deformation properties. As the research is in the initial stage, design of some measuring apparatus and establishment of

experimental techniques will also be included in this work.

The studies of toxicity of the materials and clinical behaviour of contact lenses studied herein are beyond the scope of this work and will be carried out by other members of the whole research team. However some of the clinical results will be correlated with the laboratory measurements carried out in the work.

In conclusion, the objects of the present work are two-fold:

(1) to synthesise hydrogel materials and correlate their structures and properties (and relate these to clinical behaviour) with an attempt ultimately to develop a new hydrogel contact lens for continuous wear, and

(2) to develop measuring apparatus and experimental techniques for studying some physical properties of hydrogels relevant to contact lens applications.

Chapter II : SYNTHESIS AND CHARACTERISATION OF HYDROGEL POLYMERS

This chapter contains an account of the synthesis of some hydrogel copolymers and terpolymers, together with their characterisation by density and refractive index measurements.

II.1. INTRODUCTION

As mentioned in Section I.7., polymer synthesis carried out in this work involved copolymerisation (and terpolymerisation) of some chosen vinyl monomers. Free radical copolymerisation (and terpolymerisation techniques were used in this work because of the following reasons:

(i) the vinyl monomers studied here are free-radical polymerisable,

(ii) it is easier to handle a free radical copolymerisation than an ionic or Ziegler-Natta copolymerisation, and

(iii) stereoregular copolymers are avoided because it is known in many instances that they are less oxygen-permeable than atatic copolymers synthesised by free radical copolymerisations.

Copolymers were made mainly in the form of a sheet because it was hoped that both physical property measurements and contact lenses could be made directly from each copolymer sheet. But owing to some reasons which will be explained later in this chapter, moulding lenses from a dehydrated copolymer sheet was found to be unsuccessful, and so some copolymers were later made in a rod form for lathe-cutting trials, and also in a powder form for compression moulding trials. However the copolymer hydrogel sheets were still used in physical property studies.

Some of the hydrogel copolymers thus made were characterised by their densities and refractive indices in their dehydrated state.

II.2. EXPERIMENTAL

This section involves purification of reagents and synthesis of copolymers (and terpolymers) followed by density and refractive index measurements of some of the copolymers in both hydrated and dehydrated states.

II.2.A. Reagents and Purification

In the synthesis work, the following chemicals were used which have the chemical structures shown in Figure II.1.

(1) Hydroxyethyl methacrylate abbreviated to HEMA (Mol.wt. 130)
(2) Hydroxypropyl methacrylate abbreviated to HPMA (Mol.wt. 144)
(3) Hydroxyethyl acrylate abbreviated to HEA (Mol.wt. 116)
(4) Hydroxypropyl acrylate abbreviated to HPA (Mol.wt. 130)
(5) N-vinyl pyrrolidone abbreviated to VP (Mol.wt. 111)
(6) Styrene abbreviated to St (Mol.wt. 104)
(7) para-Methyl Styrene abbreviated to pMS (Mol.wt. 118)
(8) Methyl methacrylate abbreviated to MMA (Mol.wt. 100)
(9) 4-Methyl pent-1-ene abbreviated to 4MPl (Mol.wt. 84)
(10) Ethylene glycol dimethacrylate abbreviated to EGDMA (Mol.wt. 198)
(11) Benzoyl peroxide abbreviated to BzP (Mol.wt. 242)
(12) α-Azo-isobutyronitrile abbreviated to AZBN (Mol.wt. 164)

These chemicals were purified before use. The suppliers and purification of these chemicals are shown in Table II.1. (See Appendix 2)

Reduced pressure distillations of the monomers were carried out under a stream of a nitrogen gas, and a polymerisation inhibitor (e.g. 2% hydroquinone) was added to the crude monomer before the distillation. The first portion (15%) of the distillate was always discarded and the next 50-60% of the distillate was collected and kept

in the refrigerator (with no stabiliser) for later use.

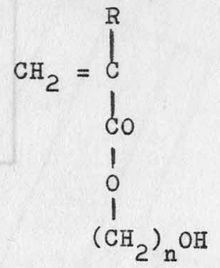
Figure II.1. Chemical structures of the reagents used

(1) HEMA (R = CH_3; n = 2)

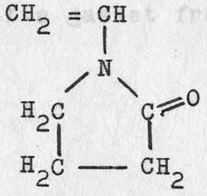
(2) HPMA (R = CH_3; n = 3)

(3) HEA (R = H; n = 2)

(4) HPA (R = H; n = 3)

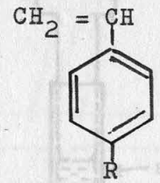


(5) N-vinyl pyrrolidone

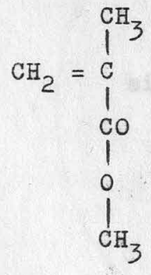


(6) Styrene (R = H)

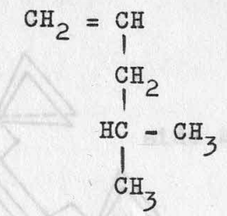
(7) para-Methyl styrene (R = CH_3)



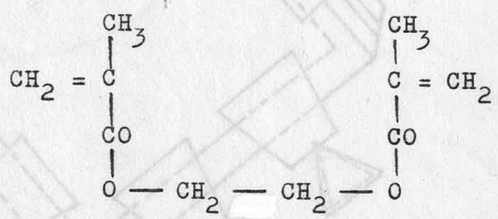
(8) Methyl methacrylate



(9) 4-Methyl pent-1-ene



(10) EGDMA



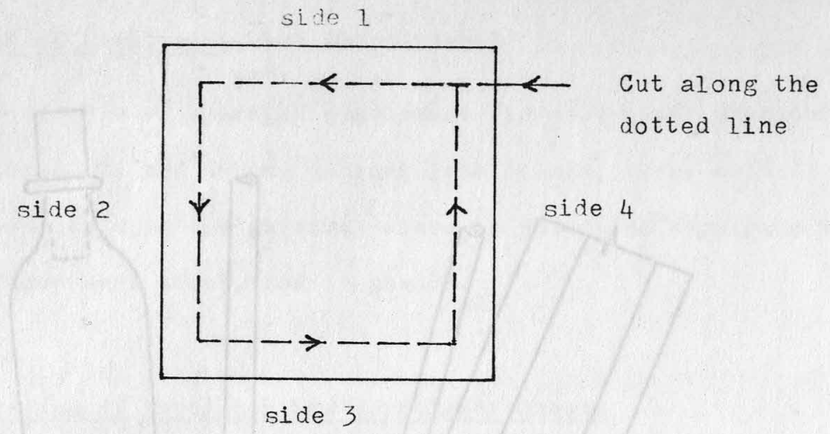


Figure II.2 Preparation of a gasket from a polythene sheet.

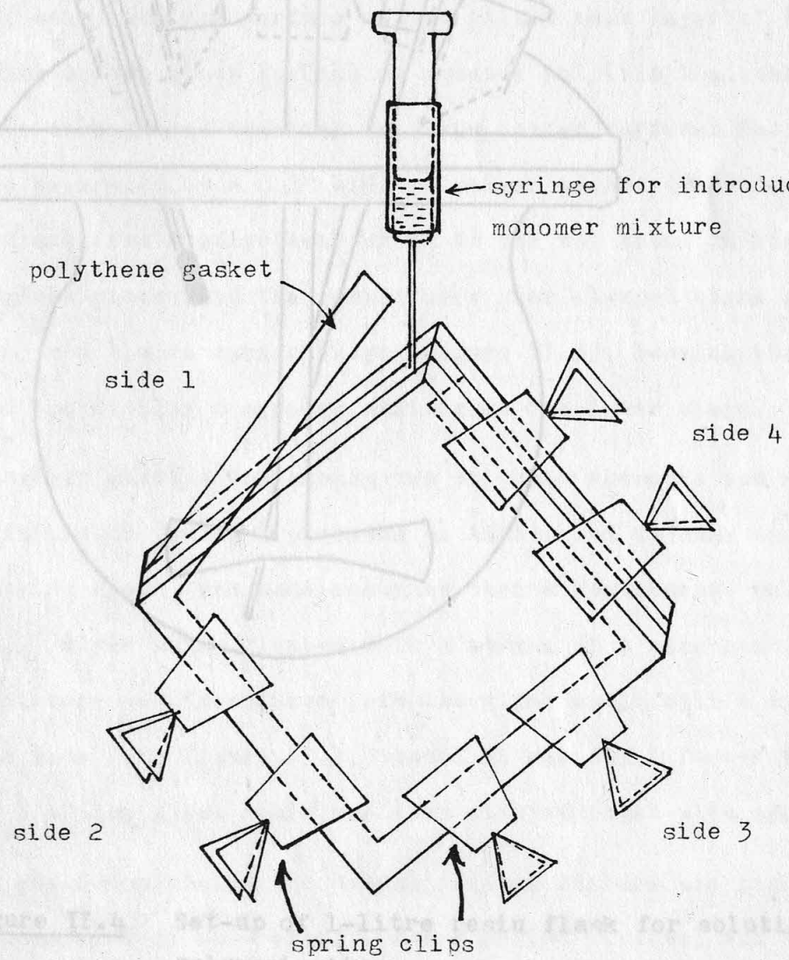


Figure II.3 A glass mould for preparing a polymer sheet.

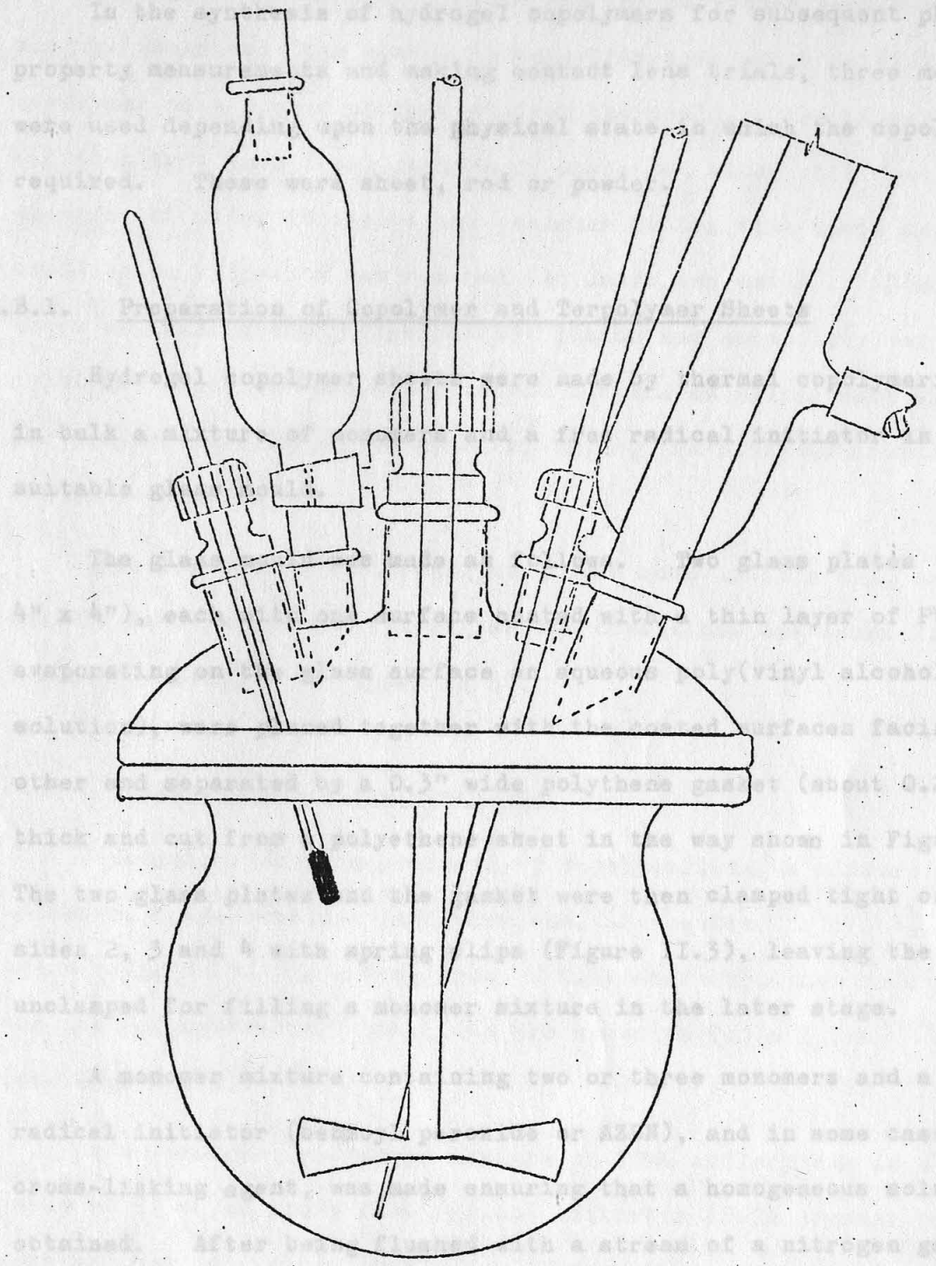
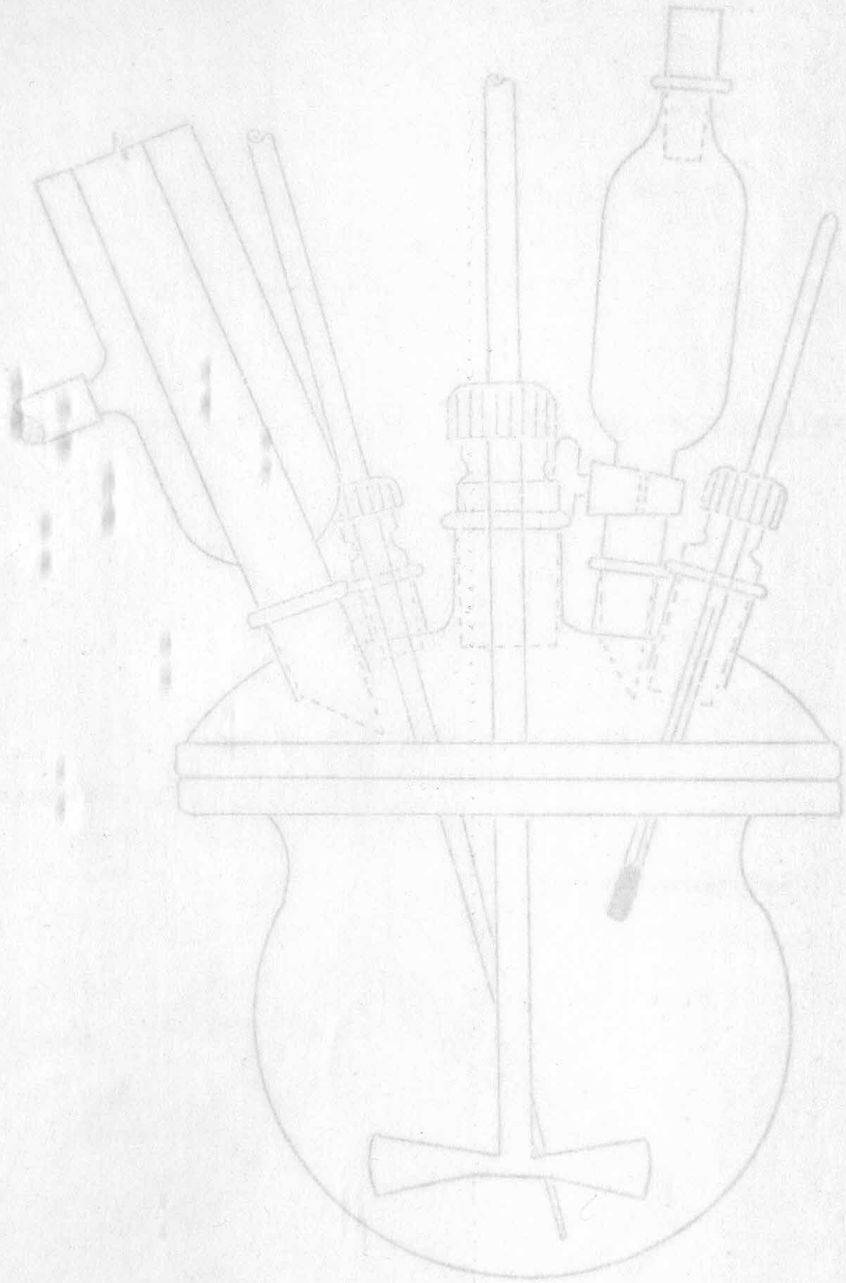


Figure II.4 Set-up of 1-litre resin flask for solution polymerisation.



II.2.B. Synthesis of Copolymers (and Terpolymers)

In the synthesis of hydrogel copolymers for subsequent physical property measurements and making contact lens trials, three methods were used depending upon the physical state in which the copolymer was required. These were sheet, rod or powder.

II.2.B.1. Preparation of Copolymer and Terpolymer Sheets

Hydrogel copolymer sheets were made by thermal copolymerisation in bulk a mixture of monomers and a free radical initiator in a suitable glass mould.

The glass mould was made as follows. Two glass plates (about 4" x 4"), each with one surface coated with a thin layer of PVA (by evaporating on the glass surface an aqueous poly(vinyl alcohol) solution), were placed together with the coated surfaces facing each other and separated by a 0.3" wide polythene gasket (about 0.2 mm. thick and cut from a polyethene sheet in the way shown in Figure II.2). The two glass plates and the gasket were then clamped tight on the three sides 2, 3 and 4 with spring clips (Figure II.3), leaving the side 1 unclamped for filling a monomer mixture in the later stage.

A monomer mixture containing two or three monomers and a free radical initiator (benzoyl peroxide or AZBN), and in some cases a cross-linking agent, was made ensuring that a homogeneous solution was obtained. After being flushed with a stream of a nitrogen gas, the monomer mixture was introduced into the glass mould with a syringe (from the side 1 in Figure II.3.) until it was about four-fifths full. The side 1 of the glass mould was then clamped tight with spring clips.

The mould containing the polymerisation mixture was placed in an oven thermostated at 60°C or 65°C for a period of time, and then in

most cases post-cured at a higher temperature. When the polymerisation was complete, all the spring clips and the polythene gasket were removed, and the mould was then soaked in water for some weeks until the copolymer sheet could be removed from the mould. The hydrated film was then kept in a fresh portion of distilled water with frequent changing of water to remove any residues in the film until an equilibrium hydration was reached (at least two weeks). Some properties (such as transparency, flexibility, fusion and solubility) of the copolymer sheet in both hydrated and dehydrated states were examined. The dehydrated film was obtained by drying the hydrated sample in a vacuum oven at 60°C.

Both copolymers and terpolymers were made in this way, and the monomer feed ratios and polymerisation conditions are shown in Tables II. 2-11. (See Appendix 2).

II.2.B.2. Preparation of Copolymer Rods

Copolymer rods were prepared by copolymerising a mixture of monomers, a free radical initiator and, if required, a cross-linking agent (e.g. EGDMA) in a test tube. Only some copolymer rods were made and the polymerisation conditions are shown in Table II.12. (see Appendix 2).

As an example, a monomer mixture of HEMA and styrene in a molar ratio of 90 to 10 and a free radical initiator (0.5% benzoyl peroxide) were weighed in a glass test tube (125mm x 16mm) about half-full. The polymerisation mixture was bubbled with a small stream of a nitrogen gas for a few minutes, and the glass tube was then closed with a suba seal. The tube was then heated with constant shaking in a water bath at 80°C for 8 hrs., and then in an oven at 90°C for 48 hrs. The copolymer

rod was removed by breaking off the glass test tube.

II.2.B.3. Preparation of Copolymer Powders (Solution Polymerisation)

To prepare uncross-linked copolymers for subsequent trials of compression moulding into contact lenses, solution copolymerisations of some comonomer pairs were carried out.

(i) Solution Copolymerisation of HEMA and 4-Methyl pent-1-ene

A mixture of HEMA, 4MPl and tetrahydrofuran was poured into a 1-litre five-neck resin flask (Figure II.4), and AZBN (0.4% by wt.) was added and dissolved. The mixture was bubbled with a small stream of a nitrogen gas for about 15 minutes and then heated at the reflux temperature for one hour. The viscous solution was precipitated in a non-solvent in a mechanical blender, and the granules thus obtained were dried in a vacuum oven at 50°C. Two attempts were made on this solution copolymerisation, and the experimental conditions are shown in Table II.13. (See Appendix 2).

(ii) Solution Copolymerisation of HEMA and Styrene in (90:10) Monomer Feed

A mixture of distilled HEMA (23.4 g.) and distilled styrene (2.1 g) [i.e. 90:10 molar ratio] with AZBN (0.1 g.) dissolved in it was added to dimethylformamide (75 ml.) in a 1-litre resin flask, and the polymerisation mixture was bubbled with a small stream of a nitrogen gas before the copolymerisation was carried out in a water bath at 70°C for 10 hrs. A viscous solution was obtained and the DMF was distilled off under vacuum at a low temperature. The residue was redissolved in methanol, and the methanolic solution was added dropwise into methyl ethyl ketone. A white precipitate was obtained and later resoaked in a fresh portion of MEK. The elastic product was torn into

small pieces, redissolved into methanol and reprecipitated as above. The final product was dried in a vacuum oven at 50°C and then ground, and the yield was about 50%.

(iii) Solution Copolymerisation of Vinyl Pyrrolidone and HPMA in (50:50) Monomer Feed

A polymerisation mixture of VP (11.1 g), HPMA (14.4 g) and AZBN (0.06 g) in methanol (40 ml) in a 1-litre resin flask was bubbled with a small stream of a nitrogen gas and refluxed at 60°C for 5 hours. The viscous solution was precipitated with water, and the rubbery precipitate was cut into small pieces, filtered and dried. It was then redissolved in methanol and reprecipitated, and the final product was dried and ground into white powder. The yield was about 40%.

(iv) Compression moulding Trials of the Copolymer Powders

Following the preparations of the linear copolymers, a preliminary investigation into thepossibility of compression moulding the copolymer powders was carried out.

The moulding apparatus (from the Kelvin Lenses Ltd.) (Plate II.1 in Appendix 6) consists of a heating block and a mould. In this work the heating block was modified with a thermostat control and with asbestos insulators surrounding the body, both to give less heat loss and less temperature fluctuation in the heating block as well as the mould during moulding. The mould of aluminium body has four cavities into which four metal discs (each has two halves with flat tops) can be inserted to allow four mouldings to be made at the same time.

The moulding trials of the copolymer powders were carried out as follows. The temperature of the heating block was preset, and one die was inserted into one cavity of the mould, which was then placed in the

heating block to preheat. When the temperature of the mould had reached about 100°C, as indicated by a thermocouple placed in one of the other three cavities of the mould, the mould was removed and the top half of the die was opened. A copolymer powder (about 0.1 g) was placed on the flat surface of the bottom half of the die, and the top half of the die was replaced in the mould, and then the mould was replaced in the heating block. A stop-watch was started, and when the temperature of the mould had reached the flow temperature of the copolymer, the vernier screw gauge on the mould was turned to load on the top half of the die, and the pressure was held there for about 3 to 60 minutes. The mould was removed from the heating block, and quenched in cold water or allowed to cool in ambient atmosphere. The die was removed from the mould and opened to release the moulded copolymer flat disc for subsequent examinations such as clarity and hydration.

II.2.C. Density Measurements of Hydrogel Copolymers

The densities of some copolymer sheets were measured in both dehydrated and hydrated states with Davenport Density Gradient Columns (Plate II.2, in Appendix 6). Three columns were filled, following the procedures mentioned in the Instruction Book for the instrument, with a mixture of two miscible liquids (e.g. carbon tetrachloride + benzene or carbon tetrachloride + heptane) giving a density range from about 1.00 to 1.32 (the densities of CCl_4, benzene and heptane are 1.595, 0.879 and 0.684 respectively). To calibrate each column, coloured marker floats of known densities were placed in the column and the positions of the floats at equilibrium were measured with a cathetometer.

For measurements of densities of the materials in the dehydrated state, samples were cut from each hydrated sheet and dried in a vacuum

oven at 60°C for several hours, and the dehydrated samples were then moistened with carbon tetrachloride and gently dropped into a suitable column of known density range. The position of the test samples were followed with the cathetometer at time intervals, until an equilibrium was reached. For measurements of densities of hydrated samples, the surface water was removed with a soft tissue paper and the test samples were moistened and dropped into an appropriate density gradient column, as in the case of dehydrated samples. Again the positions of the test samples were followed at time intervals.

The densities of HEMA - St, HEMA - MMA, HPA - St and HEMA -VP copolymers were thus measured in both dehydrated and hydrated states at 22°C. Two or three samples were used for each material, and for ease of identification, samples of different materials were cut into different shapes if they were used in the same column at the same time.

II.2.D. Refractive Index Measurements of Hydrogel Copolymers

The refractive indices of hydrated copolymer sheets were measured with an Abbé refractometer (Plate II.3). A test sample was clamped between the two prisms of the refractometer, and by adjusting the position of the light source (sodium light) as well as the knob on the side of the refractometer, a clear boundary of contrast intensities was observed at the intersection of the cross-wires in the eye-piece. The refractive index of the material was then read from the vernier scale in the refractometer.

For those test samples which did not form a good contact with the prism surfaces, a tiny drop of a suitable liquid such as benzoyl benzoate (n_D = 1.568 at 20°C)or cadmium tungstoborate Cd ($BW_{12}O_{40}$). $18H_2O$ (n_D = 1.636 at 20°C) was put on both sides of the test sample before it was clamped. For soft hydrogel sheets which contacted the

prism surfaces quite well, no contacting liquid was used.

Most of the copolymer sheets made in Section II.2.B.1 were used in this refractive index measurement at 22°C and 34°C (in the latter case, samples were hydrated at 34°C for two weeks before the measurements).

II.3. RESULTS ON POLYMER SYNTHESIS (Part 1.)

II.3.A. Purity of Monomers

All the monomers obtained from the manufacturers and commercial firms were not absolutely pure and some (e.g. HEA) were in straw colour, possibly due to the presence of a stabiliser and/or other contaminations. Distilled styrene, methyl methacrylate and vinyl pyrrolidone were found to be sufficiently pure by I.R. and G.L.C. analysis. Although reduced pressure distillation of hydroxy alkyl methacrylates and acrylates gave colourless liquids which were about 99% pure as checked by I.R. and G.L.C. analysis, a small amount of dimethacrylate or diacrylate was still found to be present in the monomer distillates, and this impurity being a cross-linking agent had a great effect on polymerisation methods, processing techniques and properties of hydrogels, as shown in the following sections. Further distillations of the hydroxy alkyl methacrylate or acrylate monomers did not seem to eliminate the dimethacrylate or diacrylate. A trial purification of hydroxyethyl methacrylate by a commercial British laboratory using a very efficient and low pressure (0.1 mm Hg) distillation apparatus and a low distillation temperature ended up with the same product (i.e. HEMA distillate with $\angle$ 1% EGDMA residue). In fact, the difficulties of obtaining a 100% pure HEMA were also encountered (and still unsolved) by the Czechoslovak workers[102] who have been using HEMA in synthesising biomedical materials including contact lens materials since 1960. They found that even if HEMA was obtained 100% pure under very careful conditions and using very efficient purification techniques, the pure HEMA would slowly, on standing at room temperature, disproportionate to EGDMA. Therefore no further effort was made in this work to obtain a 100% pure HEMA, but reduced pressure distillation of commercial HEMA

was always carried out in this work to ensure that the dimethacrylate in the HEMA distillate was not more than 1% for copolymerisation purposes. The distilled HEMA was unstabilised and kept in the refrigerator, and used as soon as possible. Similar treatments were made on the HPMA, HPA and HEA.

II.3.B. Copolymer and Terpolymer Sheets

Before a copolymer sheet was formed during bulk polymerisation, the use of a high polymerisation temperature ($> 70^{o}C$) was avoided in order to reduce the possibility of formation of bubbles in the copolymer sheet. The removal of polymer sheets by soaking the glass moulds in water was found to take one week or up to several weeks, although there was a PVA coating on each glass plate surface. A more polar and flexible material (e.g. HEA copolymers) adhered to the glass plate quite firmly and took a longer time to remove. An application of a poly(ethylene terephthalate) film to a glass plate as a mould release coating helped to ease the removal of copolymer sheets in some cases but still did not solve the mould-release problem completely.

However most copolymer (and terpolymer) sheets were made successfully, and their properties when dehydrated (e.g. transparency, rigidity, fusion, solubility) and when hydrated (e.g. transparency, rigidity) are shown in Table II.14 (See Appendix 2).

II.3.C. Copolymer Rods

Since the copolymer sheets obtained in Section II.3.B. were cross-linked even in those cases where no cross-linking agent was added, they did not melt under heat. A flat disc cut from the cross-linked sheet (dehydrated) could be moulded under heat and pressure into

a contact lens shape, but it did not retain the shape on hydration due to the relaxation of the moulding stress on hydration. Therefore some copolymer rods were made so that they could possibly be lathe-cut into contact lenses - the conventional contact lens manufacturing process.

Some of the experimental results on polymer rods are shown in Table II.15 (see Appendix 2).

In the preparation of HEMA-St (90;10) rods, the surfaces of the rods were found to contain small irregular "drains" probably due to contraction of the copolymer from the glass test tube during polymer and network formations. However such surface irregularities could be polished off during lathe-cutting, and the techniques of lathe-cutting trials are out of the scope of this thesis.

Although HEMA-St (90:10) copolymer rod was removed with no difficulties from the glass test tube (by breaking it), HPA-VP (80:20) rod was found to adhere firmly to the glass test tube and present a severe mould-release problem. A recent development in the research group of easing out the mould-release problem is to use a polythene tube for the preparation of polymer rods. Preparation of a large quantity of polymer rods for lathe-cutting trials is being carried out by another member of the research team and is out of the scope of this thesis.

The rod with malonic acid in it was cut in a lens-shaped blank and then compression moulded into a lens, with the hope that malonic acid would be activated during the processing conditions to give additional cross-links which would then retain the lens shape. But this was found to be unsuccessful, as the lens returned to its original state on subsequent hydration, illustrating again that a copolymer once it was cross-linked could not be compression moulded into a successful contact lens.

The main difficulty in preparing a perfect polymer rod is to get it completely bubble-free, since bubbles in the polymer rod will affect the lathe-cutting techniques and quality of the lathe-cut contact lenses. The bubbles were found to form mainly when gelation started, and the possibility of such bubble formation was found to be higher in the case of polymer rods than in the case of polymer sheets. However the bubble formation in the polymer rods could be avoided by (a) preparing the rods in a water bath rather than in an oven, (b) using a lower initiator concentration and a lower polymerisation temperature although this means that a longer polymerisation time was required, and (c) gently and frequently shaking the polymerisation mixture before the gelation occurred.

II.3.D. Copolymer Powders

It is interesting to see if a contact lens could be compression moulded from a polymer powder perhaps with cross-linking occurring during moulding (just like vulcanising rubber during moulding). In order to flow and be moulded under heat and pressure, a polymer must be uncross-linked or linear in structure. To prepare a linear copolymer, solution copolymerisation was carried out.

Since poly(4-methyl pent-1-ene) has been known[199] to have a high oxygen permeability, it was hoped that a copolymer of HEMA-4MP1 would have a higher oxygen permeability than PHEMA. Solution copolymerisation of HEMA and 4MP1 in THF under the experimental condition (A) [See Table II.13 in Appendix 2] was found to give a partially cross-linked product, which could not be fused completely under heat but could be pressed at 240°C to give a translucent disc which subsequently lost its dimensional stability on hydration. Under the experimental condition (B), the product obtained was transparent and could be pressed into discs which on hydration lost their shapes and became sticky (sign of solubility).

These observations suggest that in solution copolymerisation of HEMA, linear copolymers could be made by using a larger proportion of the comonomer (e.g. 4MP1) and/or solvent in the copolymerisation mixture. But as the solvent concentration was increased, it was more likely to yield a lower molecular weight copolymer. Infra-red analysis of the copolymerisation product did not give a conclusive indication that 4MP1 had been incorporated into the structure. But on the other hand, solution homopolymerisation of 4MP1 alone in THF (10 mls each) gave a greasy product (very low yield < 0.1 g) having an I.R. spectrum similar to poly(4-methyl pent-1-ene). So the product obtained in the solution copolymerisation of HEMA and 4MP1 might contain only a small amount, if any, of 4MP1 units in its structure, probably due to the lack of radical stabilising factor in the 4MP1 unit during propagation steps of the copolymerisation. An extension to the work on the copolymerisation of HEMA and 4MP1 carried out by Cobbett[201] and Beerman[202] using a complex copolymerisation technique developed by Gaylord[203], i.e. copolymerising HEMA and 4MP1 in the presence of a Friedel-Crafts halide or a Lewis acid such as zinc chloride and optionally a free radical initiator such as AZBN, did not result in a copolymer containing a substantial amount of 4MP1 units.

Solution copolymerisation of a monomer mixture of HEMA and styrene in 90:10 molar ratio gave a white powder, soluble in methanol, dimethyl-formamide and dimethylsulphoxide, but insoluble in carbon tetrachloride and chloroform. This HEMA-St copolymer (checked by I.R. analysis) melted under heat and could be compression moulded into thin discs. A preliminary investigation into moulding conditions and an addition of a cross-linking agent (e.g. succinic anhydride) was made on the HEMA-St (90:10) and the results are shown in Table II.16 (See Appendix 2).

Solution copolymerisation of a monomer mixture of vinyl pyrrolidone and HPMA in (50:50) ratio also gave a white powder which was shown by I.R. analysis to be VP-HPMA copolymer. The copolymer was fusible when heated and soluble in methanol, DMF, DMSO, but insoluble in petroleum ether, carbon tetrachloride. A preliminary moulding trial of this VP-HPMA (50:50) copolymer was also investigated and the results are also shown in Table II.16 (see Appendix 2).

In almost all cases, the dehydrated discs were transparent except in a few cases where the copolymer or the cross-linking agent was not completely fused (e.g. low moulding temperature). But on hydration, these discs became translucent or opaque, and this was thought to be due to phase separation of water from the polymer probably arising from (a) incomplete fusion of the particles during moulding, and this might be related to moulding conditions, moulding techniques and linearity of the copolymer, (b) presence of some styrene-rich chains in the HEMA-St copolymer, and (c) absorption of too much water by the copolymer on hydration, as an incorporation of a cross-linking agent into the moulding powder or use of a 2% saline solution for hydration was found to reduce the loss of transparency in the hydrated samples.

Another difficulty during the moulding was the mould-release problem. The polar copolymer adhered to the die (quite severely in some cases), resulting in broken pieces as the moulded disc was removed from thetool. Coating the die surface with a thin layer of PVA before moulding helped to solve the mould-release problem to some extent but not completely.

Further developments on the moulding techniques, moulding conditions and mould release were carried out by another member of the research team and his results are beyond the scope of this thesis.

II.4. RESULTS ON POLYMER CHARACTERISATION (Part 2)

II.4.A. Densities of some Hydrogel Copolymers

In a density gradient column, the density of the liquid mixture increases from the top to the bottom, but it was found in the calibration of the column that the density might not be linear with the height of the liquid column (i.e. with the cathetometer reading) over the whole range. An example of this non-linearity is shown in Figure II.5, where the density of the column liquid was plotted vs. the cathetometer reading of the marker floats. This may be due to the difficulties, during the column preparation, in controlling a proper rate of mixing of the two liquids and a proper flow rate of the liquid mixture into the column. Also the density gradient in the column was sometimes found to change if the column liquid mixture was greatly disturbed (e.g. due to lifting and inserting the sweep basket during the process of removing samples from the column after an experiment), or if the column liquid mixture was aged (e.g. due to diffusion of one part of the liquid mixture into another, or due to impurities leached out from test samples into the column liquid). However a calibration must be carried out before any density measurement was made. When a marked change in the density gradient was found in the column, a new column was remade.

The density of a material was obtained by taking the cathetometer reading of the sample at equilibrium in the column and reading the corresponding density from the density calibration curve. Since it was found that an equilibrium was hard to obtain in some cases, the position of the sample in the column (i.e. the cathetometer reading and hence the density) was followed at some time intervals. This was carried out with both hydrated and dehydrated samples, and some examples are shown in Figures II.6 and II.7 respectively. It can be seen from Figures II.6 &

Figure II.5 The calibration curves of two density gradient columns A and B made up of carbon tetrachloride and benzene mixture. (A_2 is the new calibration curve of the column A one week later than A_1).

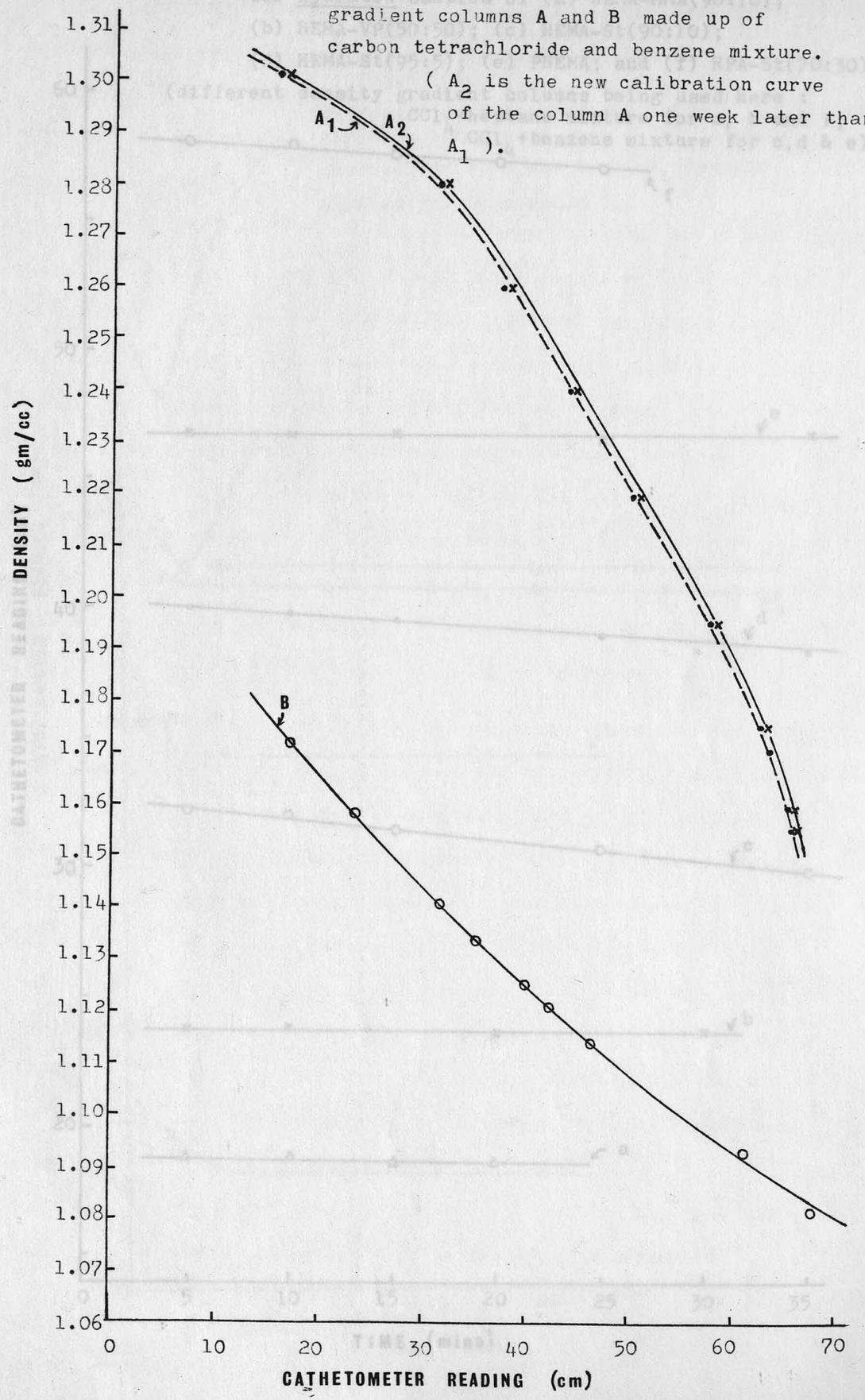
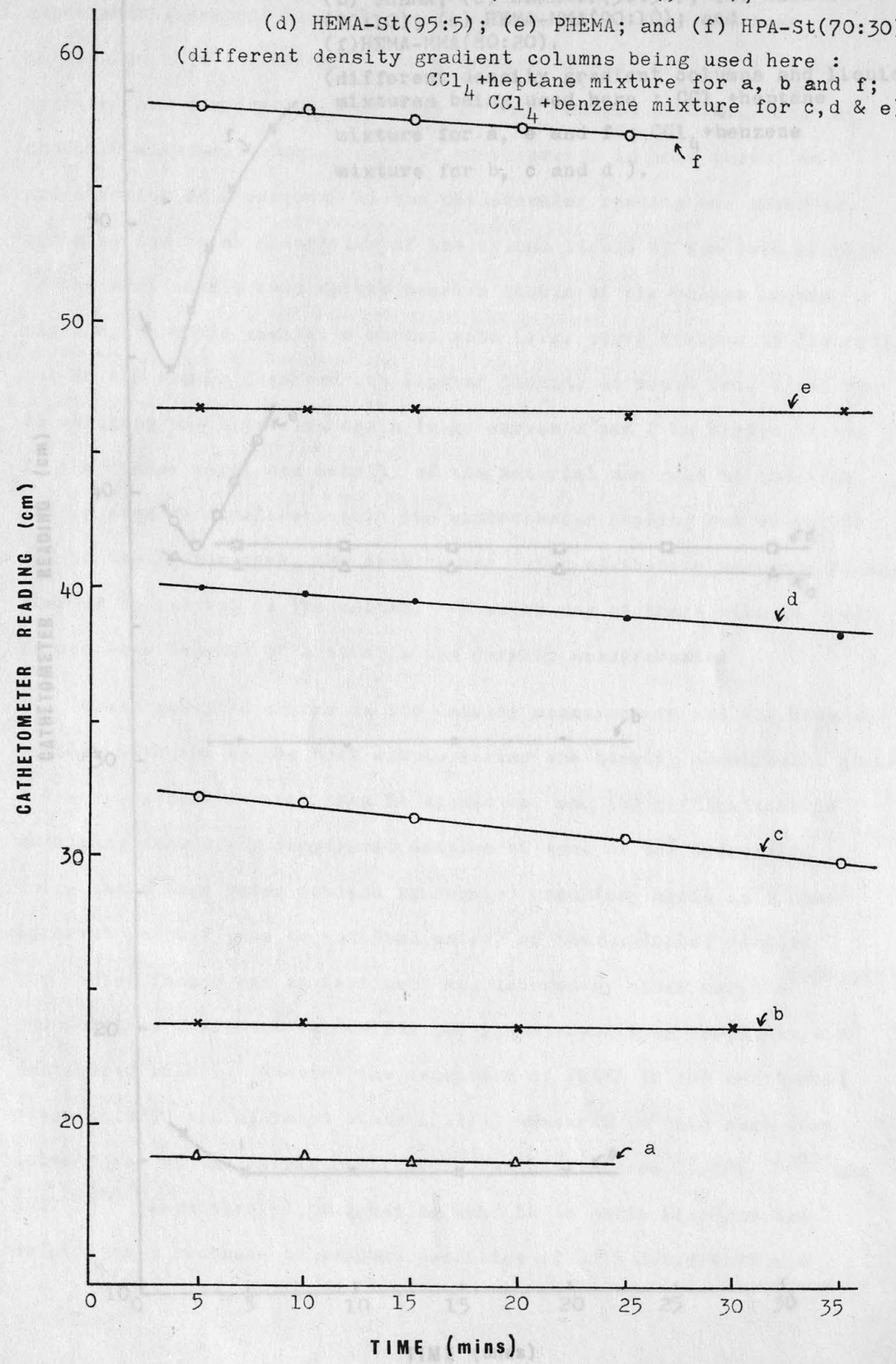


Figure II.6 Dependence of cathetometer readings on time for hydrated samples of (a) HEMA-MMA(90:10); (b) HEMA-VP(50:50); (c) HEMA-St(90:10); (d) HEMA-St(95:5); (e) PHEMA; and (f) HPA-St(70:30). (different density gradient columns being used here : CCl_4+heptane mixture for a, b and f; CCl_4+benzene mixture for c,d & e)



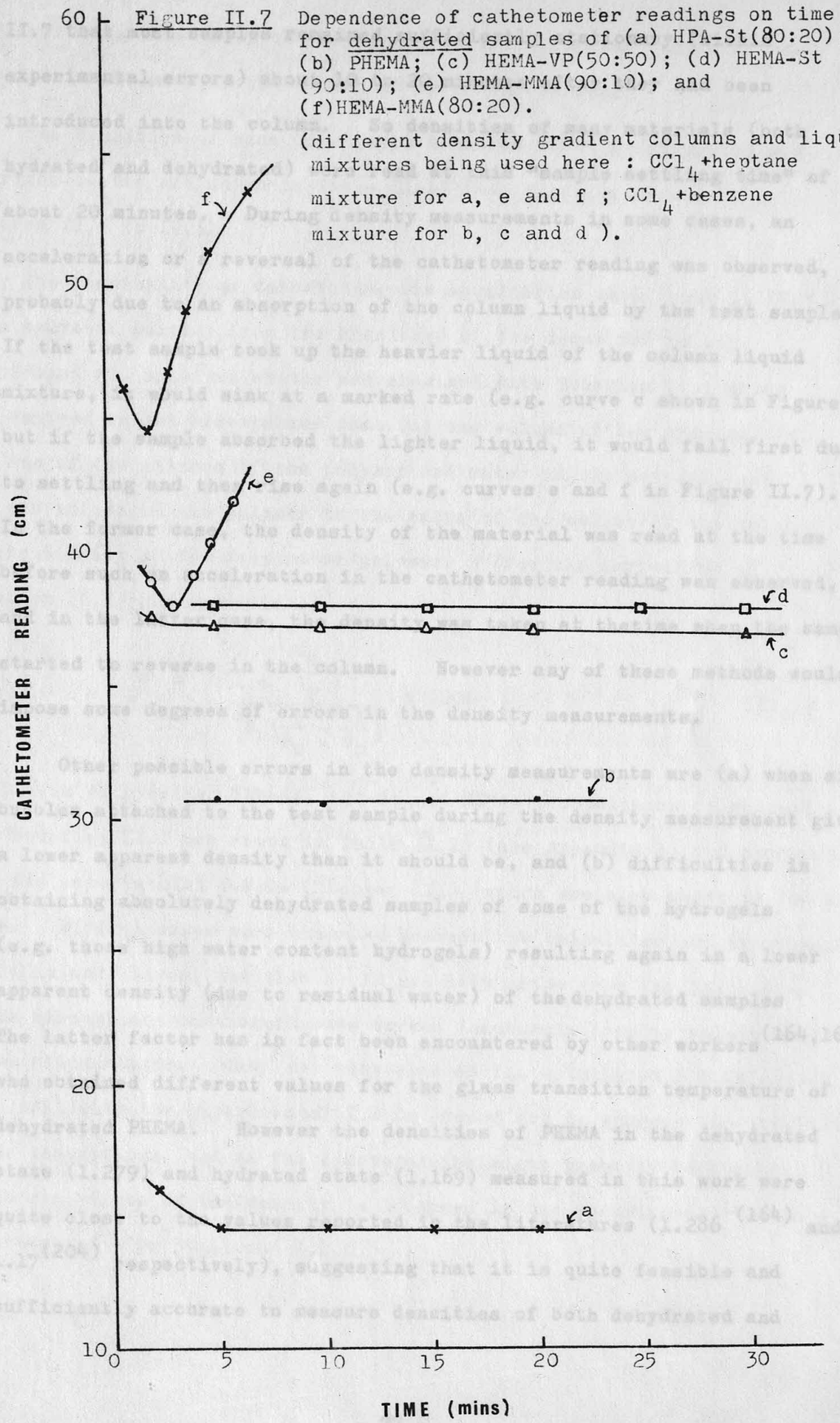
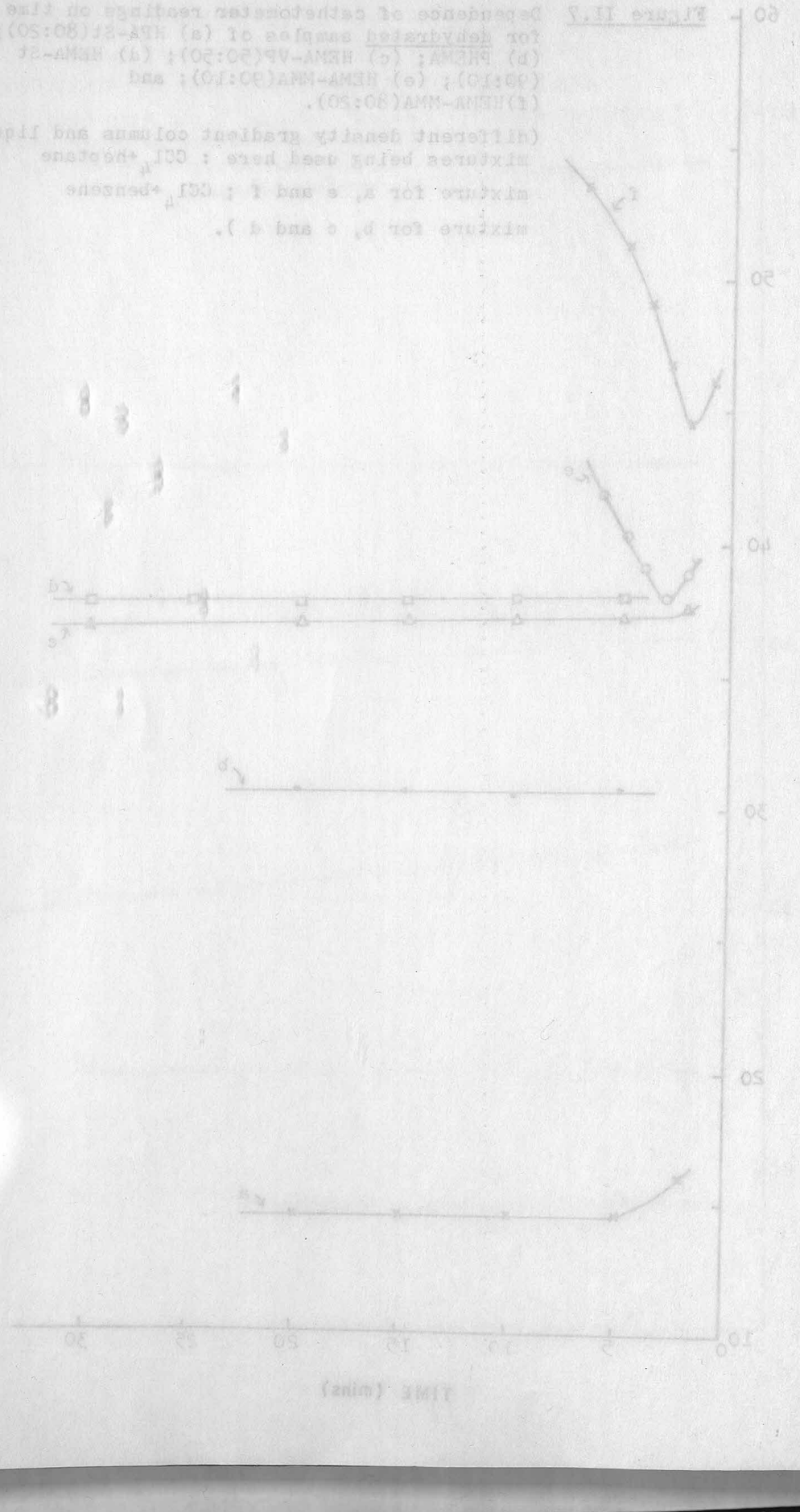


Figure II.7 Dependence of cathetometer readings on time for dehydrated samples of (a) HPA-St(80:20); (b) PHEMA; (c) HEMA-VP(50:50); (d) HEMA-St (90:10); (e) HEMA-MMA(90:10); and (f)HEMA-MMA(80:20).

(different density gradient columns and liquid mixtures being used here : CCl_4+heptane mixture for a, e and f ; CCl_4+benzene mixture for b, c and d).



II.7 that most samples remained sufficiently stationary (within experimental errors) about 10 to 20 minutes after they had been introduced into the column. So densities of many materials (both hydrated and dehydrated) were read at this "sample settling time" of about 20 minutes. During density measurements in some cases, an acceleration or a reversal of the cathetometer reading was observed, probably due to an absorption of the column liquid by the test samples. If the test sample took up the heavier liquid of the column liquid mixture, it would sink at a marked rate (e.g. curve c shown in Figure II.6) but if the sample absorbed the lighter liquid, it would fall first due to settling and then rise again (e.g. curves e and f in Figure II.7). In the former case, the density of the material was read at the time before such an acceleration in the cathetometer reading was observed, and in the latter case, the density was taken at thetime when the sample started to reverse in the column. However any of these methods would impose some degrees of errors in the density measurements.

Other possible errors in the density measurements are (a) when air bubbles attached to the test sample during the density measurement giving a lower apparent density than it should be, and (b) difficulties in obtaining absolutely dehydrated samples of some of the hydrogels (e.g. those high water content hydrogels) resulting again in a lower apparent density (due to residual water) of the dehydrated samples The latter factor has in fact been encountered by other workers[164,165] who obtained different values for the glass transition temperature of dehydrated PHEMA. However the densities of PHEMA in the dehydrated state (1.279) and hydrated state (1.169) measured in this work were quite close to the values reported in the literatures (1.286[164] and 1.17[204] respectively), suggesting that it is quite feasible and sufficiently accurate to measure densities of both dehydrated and

hydrated materials using the Davenport Density Gradient Column if the liquid mixture used in the column is correctly chosen.

The densities of some copolymers measured in both dehydrated and hydrated states are shown in Table II.17 (see Appendix 2) and Figures II.8 to II.11.

The possibility of calculating the equilibrium water content (E.W.C.) of a hydrated polymer from the knowledge of the densities in both dehydrated and hydrated states was examined with Equation II.1 which was derived on two assumptions that (a) the volume of the hydrogel is the sum of the volumes of the polymer and water in the network, and (b) the volume of the polymer is the ratio of the weight of the polymer to the density of the dehydrated polymer. (For the derivation of Equation II.1, see Appendix 1).

$$\frac{W_W}{W_G} = \frac{(1 - \frac{d_G}{d_P})}{d_G\,(\frac{1}{d_W} - \frac{1}{d_P})} \times 100\% \qquad \text{(II.1)}$$

The E.W.Cs of hydrated HEM-St, HPA-St and HEMA-VP copolymers calculated from Equation II.1 are shown in Table II.18 (see Appendix 2) and compared with the experimental E.W.Cs (Chapter III) which are also shown in the same table. Discrepancies were observed between the calculated and experimental E.W.Cs in some cases, and this could possibly arise from the invalidity of the assumptions mentioned above or/and inaccurate density values used in the calculations. The fact that some of the calculated E.W.Cs agreed quite well with the experimental E.W.Cs seemed not to approve the invalidity of the assumptions, and so the discrepancies might probably be due to inaccurate values of the density (a 1% error in the density resulted in a difference of 4 or 5% in the E.W.C.).

II.4.B. Refractive Indices of some Hydrogel Copolymers

During measurements of refractive indices of the materials with an Abbe refractometer, the main difficulty was in obtaining a good contact between the test sample and the prism surfaces (i.e. a clear boundary when viewed through the eyepiece). It was found that during the refractive index measurements of dehydrated sample sheets and of those non-gel hydrated samples, a contacting liquid was necessary, and the choice of benzoyl benzoate or cadmium tungstoborate as the contacting medium must satisfy a requirement that the refractive index of the contacting liquid must be higher than that of the test sample, so that two boundaries of different intensities could be observed in the dark field of the eyepiece, one being the refractive index of the contacting liquid and the other that of the sample. For hydrogels no contacting liquid was required because the contacts between the test sample and the prism surfaces were quite good, and it was observed that the boundary of contrast intensities was quite clear in some cases and diffused in other cases, but still distinguishable. So the uncertainty in reading the boundary would impose some degrees of inaccuracy in the refractive indices measured.

However the refractive indices of some copolymers measured in both dehydrated and hydrated states at 22°C and 34°C are shown in Table II.19 (see Appendix 2). The refractive indices of dehydrated and hydrated PHEMA at 22°C were found in this work to be 1.512 and 1.435 respectively, which are quite consistent with the literature values[204] of 1.51 and 1.4

The possibility of calculating the equilibrium water content (E.W.C.) of a hydrogel from its refractive indices in both dehydrated and hydrated states was attempted using Equation II.2, the derivation of which can be seen in Appendix 1. The calculated E.W.Cs of hydrated HEMA-St copolymers

at 22°C were found to be quite close to the experimental E.W.Cs with a deviation of about 1% in most cases, as shown in Table II.20 (see Appendix 2).

$$\frac{W_W}{W_G} = \frac{\dfrac{n_G^2 - 1}{n_G^2 + 1} - \dfrac{n_P^2 - 1}{n_P^2 + 2}}{d_P\left[\dfrac{n_W^2 - 1}{n_W^2 + 2} - \dfrac{n_G^2 - 1}{n_G^2 + 2}\right] + \left[\dfrac{n_G^2 - 1}{n_G^2 + 2} - \dfrac{n_P^2 - 1}{n_P^2 + 2}\right]} \qquad \text{(II.2)}$$

II.5. DISCUSSIONS

The experimental results on the flexibility, optical clarity, densities and refractive indices of the copolymer sheets in both dehydrated and hydrated states will be discussed here.

In almost all cases (as shown in Table II.14) where a hydroxy alkyl methacrylate or acrylate was one of the copolymer components, the dehydrated copolymers were found to be infusible when heated and insoluble in any solvent, suggesting that the copolymers were in a cross-linked state. Although no cross-linking agent (EGDMA) was added to the polymerisation mixture, cross-linking of the copolymers and terpolymers did occur during polymerisation step and probably arose from chain transfer[128] as well as from the residual dimethacrylate or diacrylate in the monomer which could not be completely removed as mentioned in Section II.3.A. However those copolymers containing only 10% HEMA or HPA or HEA were found to be fusible and soluble in some solvents (e.g. benzene).

In the dehydrated state, many homopolymers (such as PHEMA and PHPMA) and copolymers (such as HEMA-St, HEMA-MMA, HEMA-VP and HPMA-VP pairs) over the whole composition range were found to be as rigid and brittle as PMMA

and PS (i.e. polystyrene), but the corresponding styrene copolymers containing HPA or HEA were comparatively flexible, and the flexibility increased as the HPA or HEA content in the copolymer was increased (Table II.14 in Appendix 2). This can be explained by the restriction of main chain rotation in the HEMA and HPMA copolymers due to the presence of the methyl group on the main chain, as analogous to the rigidity of poly(methyl methacrylate) compared with poly(ethyl acrylate) or poly(methyl acrylate).

When hydrated, the copolymers became flexible or remained sufficiently rigid depending on the amount of water absorbed and on the interaction between water molecules and the polar chains of the copolymers. Like the effect of a plasticiser on poly(vinyl chloride), it is very likely that water molecules in a hydrogel plasticised the polymer and made the hydrogel flexible. Hydrated HEMA-St, HPMA-St, HEA-St and HPA-St copolymer series with various compositions, changed continuously from a rigid film to a semi-rigid and then to a flexible film as the content of the polar unit (i.e. HEMA, HPMA, HEA or HPA) was increased, i.e. as the hydrophilicity (or equilibrium water content) of the copolymer increased. Most of the other hydrogels (sheets 39 to 62 in Table II.14) were flexible and apparently they contained a large amount of water in the networks. The flexibility of a dehydrated copolymer series seemed to affect the copolymer composition at which such a change of its hydrated materials from "rigid" to "flexible" took place, as suggested by the results shown in Table II.14, in which both the hydrated HEA-St and HPA-St copolymers became semi-rigid when they had about 50 to 60% HEA and HPA in the copolymers respectively, whereas hydrated HEMA-St copolymers became semi-rigid only when the copolymer had about 80% HEMA. The results that when hydrated, VP-St (20:80) and

(30:70) copolymers were weak and had no film properties compared with the rigid HEMA-St or HPA-St copolymers at the same compositions, and that cross-linked poly(vinyl pyrrolidone) hydrogels (sheets 44 and 45) were weaker and less flexible than the HPA-VP-St (35:55;10) hydrogel (Sheet 60) having about the same water content suggest that water has less plasticising effect on (or less interaction with) vinyl pyrrolidone than with the hydroxyl containing groups such as HPA and HEMA.

Although glass transition temperatures (T_g) of the hydrated copolymers were not measured due to experimental difficulties, apparently those flexible hydrogels had T_g lower than the room temperature at which rigidity and flexibility were assessed, and those rigid hydrated copolymers had T_g higher than the room temperature, and the semi-rigid hydrated copolymers had T_g somewhere near the room temperature.

Table II.14 also shows that most of the copolymers (and terpolymers) were optically clear when dehydrated, suggesting that they were homogeneous at least visually. Incompatibility was observed in some copolymers such as VP-St(30:70). Equilibrium hydration of those transparent dehydrated copolymer sheets resulted in some cases optically clear materials and in other cases translucent or opaque materials. Hydrated copolymers of HEMA-St, HPMA-St, HPA-St and HEMA-MMA were found to be transparent over the composition range studied, except hydrated HPA-St (60;40) copolymer which was found to be translucent. It could be that at this composition, the polymer chains of hydrated HPA-St(60:40) were in such a configuration resulting from the plasticisation effect of water, that clusters of water molecules were present in the copolymer network giving rise to a microphase separation of different refractive indices. HPMA-VP(50:50) hydrogel was found to be slightly translucent, but incorporation of an additional cross-linking agent (EGDMA) resulted in a transparent hydrogel (sheets 47

and 48), probably because of a tighter network structure and a reduction of water absorption at equilibrium hydration. The Table also shows that the VP-St copolymer series containing more than 10% VP gave an opaque material on hydration whereas hydrated HEMA-St, HEA-St, etc. copolymers were transparent at the same composition. It is very likely that the VP-St copolymer was a mixture of vinyl pyrrolidone-rich copolymer and styrene-rich copolymer arising from the large difference in reactivity ratios of VP and St (0.045 and 15.7 respectively)[200] in the copolymerisation, and that water molecules had a weaker interaction with VP than with HEMA or HEA, resulting in the formation of water clusters in the hydrated VP-St copolymer having different refractive indices. The latter reasoning was also supported by the results obtained in which cross-linked poly(vinyl pyrrolidone) hydrogel (sheets 44 and 45) were translucent and had poorer strength (torn more easily), compared with the transparent and stronger hydrogel of HPA-VP-St (35:55:10) having about the same water content.

Infra-red spectra of some of the hydroxy containing copolymer sheets in the dehydrated state were made and found to contain broad bands. So the copolymer sheets were characterised by their densities and refractive indices. In the dehydrated state, the densities of the four copolymer series were quite linear with the compositions of their monomer feeds (as shown in Figures II.8 to II.11). The deviations of some of the points from the straight lines could arise from (a) the difficulties in obtaining the true densities of the copolymers due to the possible errors mentioned above, and (b) that the composition of a copolymer might not be the same as the composition of its monomer feed, resulting from the difference in reactivity ratios of the monomers in the copolymerisation or from the experimental conditions (e.g. percentage conversion). However such

deviations seemed to be less in the HEMA-St and HPA-St copolymer series, suggesting that HEMA and HPA had similar reactivity ratio as styrene in the copolymerisations. The literature values[205] of the reactivity ratios of HEMA and Styrene (0.65 and 0.57 resepctively) support our results.

It is interesting to see in Figures II.8 to II.10 that the correlation between densities of the hydrated samples of a copolymer series and its compositions (such as HEMA-St, HPA-St or HEMA-MMA pairs) passed through a maximum density at a copolymer composition, which depended on the chemical structure of the copolymer. The maximum densities of the hydrated samples were found to be at about 75% HEMA, 60% HPA and 45% HEMA in the HEMA-St, HPA-St and HEMA-MMA copolymers respectively. These compositions at which maximum densities of the hydrated samples were obtained are quite consistent with the compositions at which the transition of the hydrated samples from "rigid" to "flexible" took place (Table II.14). So, as the content of the polar (hydrophilic) group in a copolymer was increased from zero, the increasing amount of water absorbed started to cause neighbouring copolymer chains to separate slowly but virtually did not affect the density of the dehydrated copolymer until such a copolymer composition was reached at which the chain separation became obvious resulting in a marked change in water absorption, and the density of the hydrated copolymer started to reverse and decreased (in contrast to the case of the dehydrated copolymer) as the polar group content was further increased because water has a lower density than either of the homopolymers. This phenomenon of a maximum density in the hydrated samples of a copolymer having various compositions, together with the observation of flexibility and rigidity in those materials lead to a suggestion that the copolymer composition at which

maximum density of a hydrated copolymer series is obtained can also be the composition separating non-hydrogels and hydrogels for that copolymer series.

At equilibrium hydration, HEMA-VP copolymers at all compositions are hydrogels since they are flexible and the densities of the hydrated samples decreased at a faster rate than those of the dehydrated samples as the VP content in the copolymer was increased (Figure II.11) which also suggests that VP is more hydrophilic than HEMA.

As shown in Figure II.12, the refractive indices of dehydrated HEMA-St copolymers decreased linearly with increasing HEMA content in the copolymer compositions, supporting the conclusion made from the density measurements of the samples that these copolymers had about the same composition as their monomer feed compositions. For hydrated HEMA-St copolymers, the refractive index decreased markedly at the copolymer composition of about 75% HEMA, substantiating again that a real hydrogel of HEMA-St copolymer started at this composition because water has a lower refractive index (1.333) than either PHEMA or polystyrene.

The correlation between refractive indices of some hydrated copolymers and their equilibrium water contents (E.W.C.) (determined experimentally in Chapter III) at 22°C and 34°C is shown in Figure II.13, from which it can be seen that refractive indices of hydrated copolymers having less than 30% E.W.C. were more dependent on the copolymer structure than those hydrogels having more than 30% E.W.C., and that the temperature effect on the refractive indices of the hydrogels seems to arise mainly from the temperature effect on theis E.W.Cs.

The effect of the equilibrium water content of a hydrogel on its refractive index was also examplified by the HPMA-VP (50:50) and HPMA-VP (80:20) hydrogels having various E.W.Cs controlled by various degrees of cross-linking, and is shown in Figure II.14. The refractive index of

the hydrogels was found to decrease by about 0.0021 for every increase of 1% E.W.C. at 34°C.

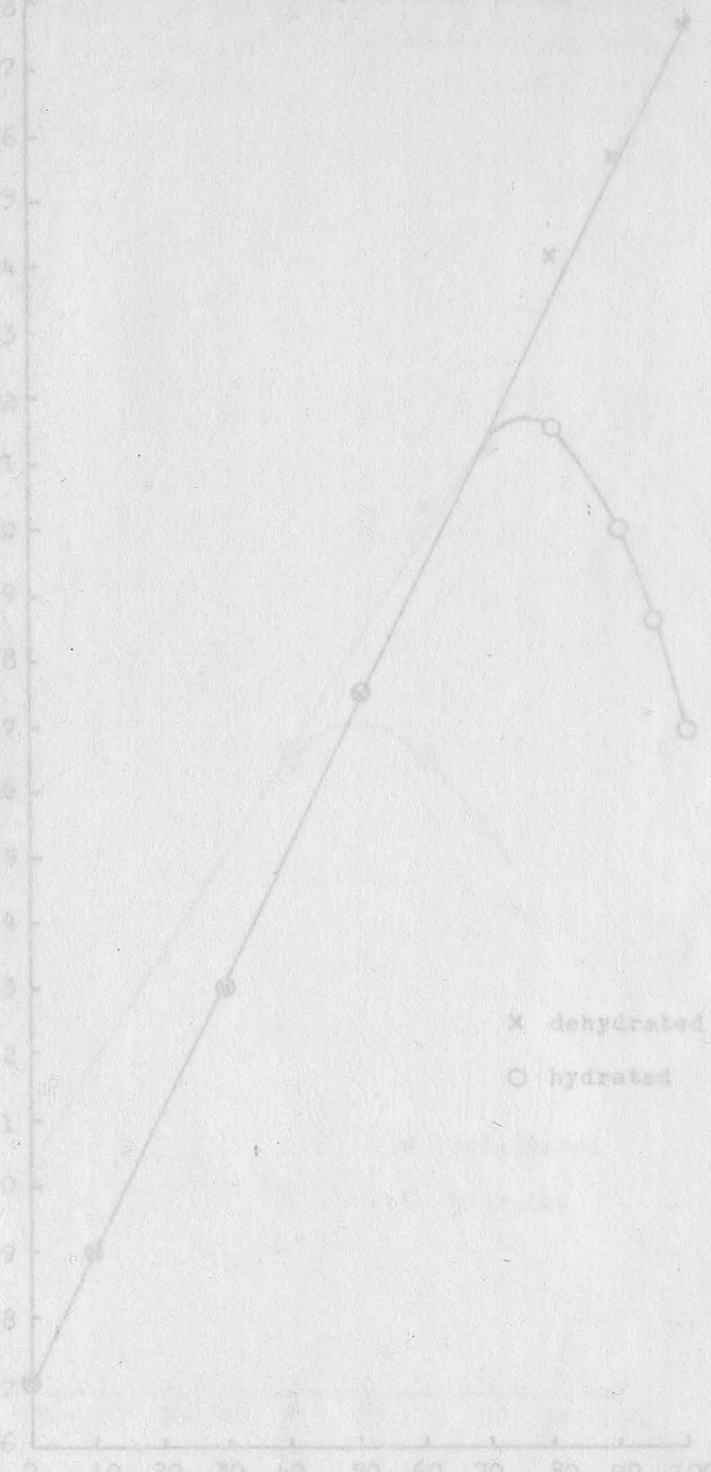


Figure II.8 Densities of hydrated and dehydrated HEMA-St copolymers of various compositions at 22°C.

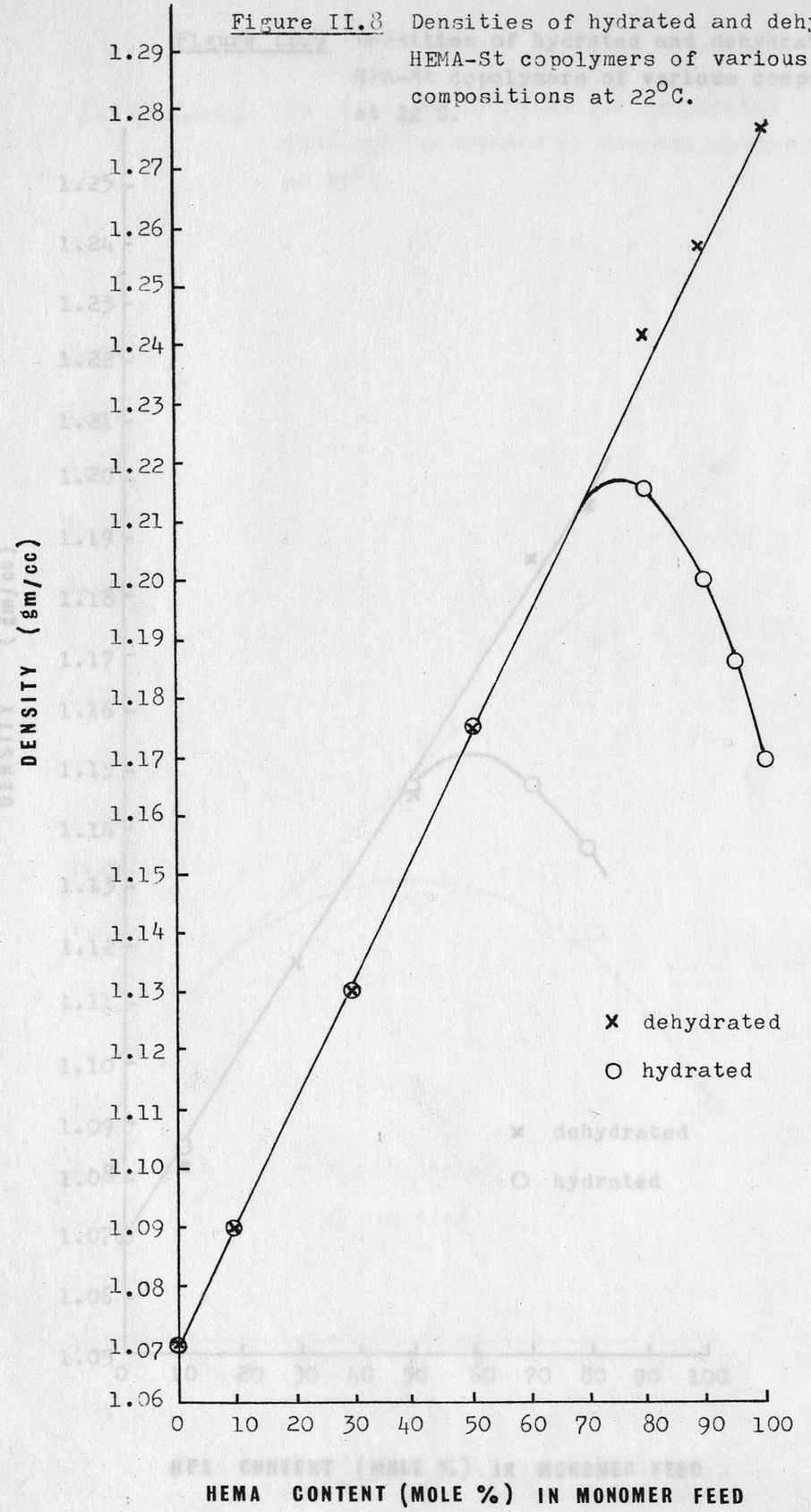


Figure II.9 Densities of hydrated and dehydrated HPA-St copolymers of various compositions at 22°C.

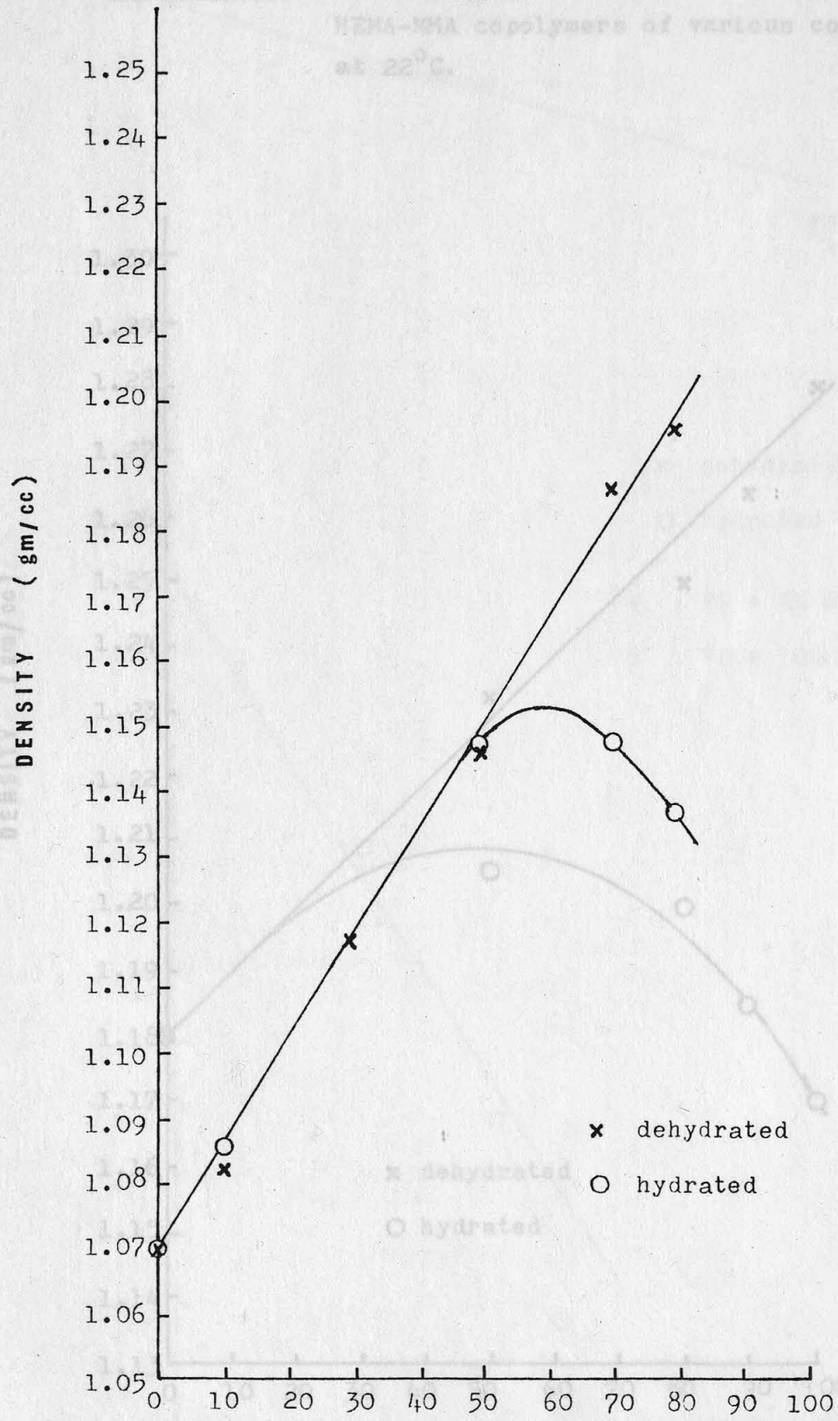


Figure II.10 Densities of hydrated and dehydrated HEMA-MMA copolymers of various compositions at 22°C.

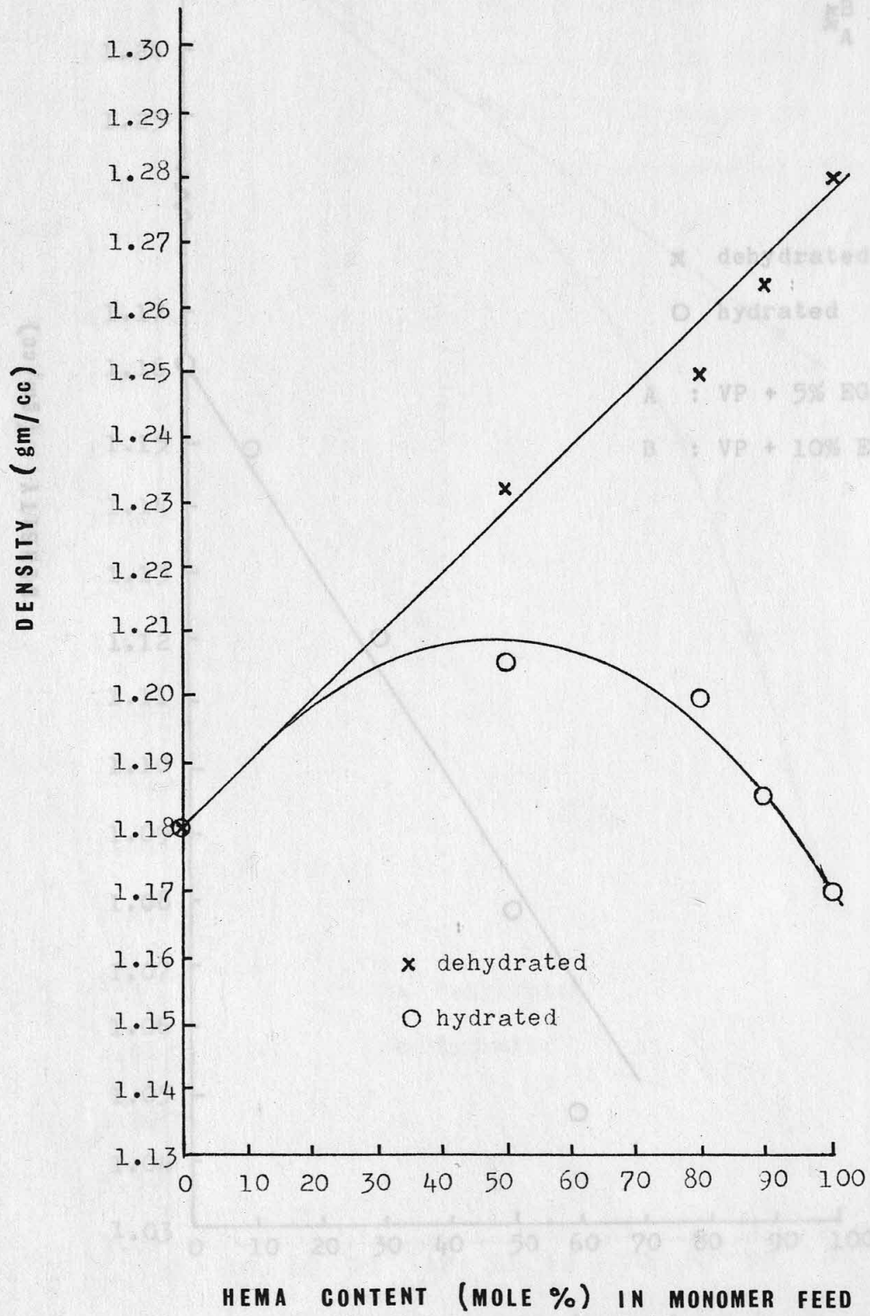


Figure II.11 Densities of hydrated and dehydrated HEMA-VP copolymers of various compositions at 22°C.

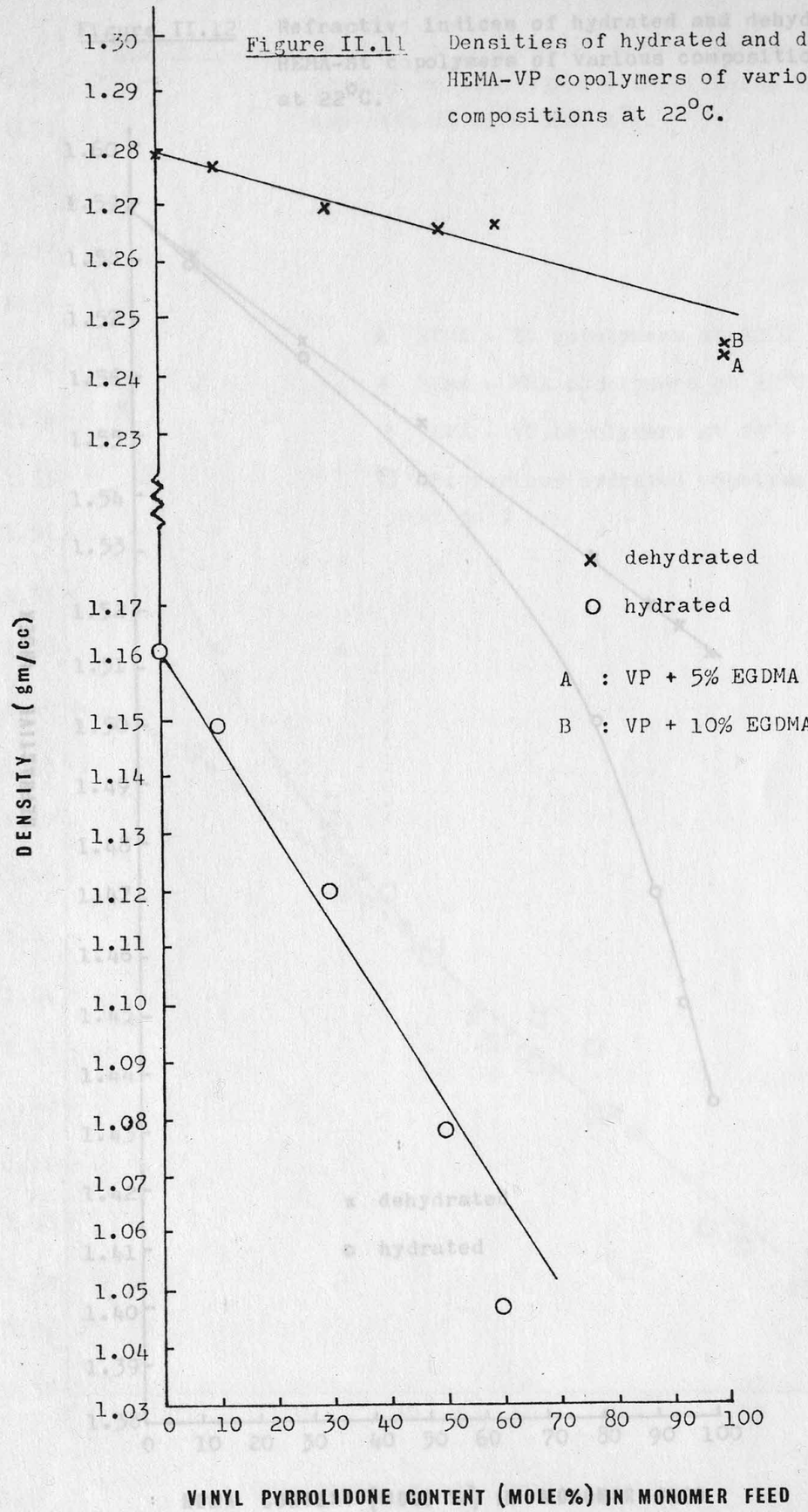


Figure II.12 Refractive indices of hydrated and dehydrated HEMA-St copolymers of various compositions at 22°C.

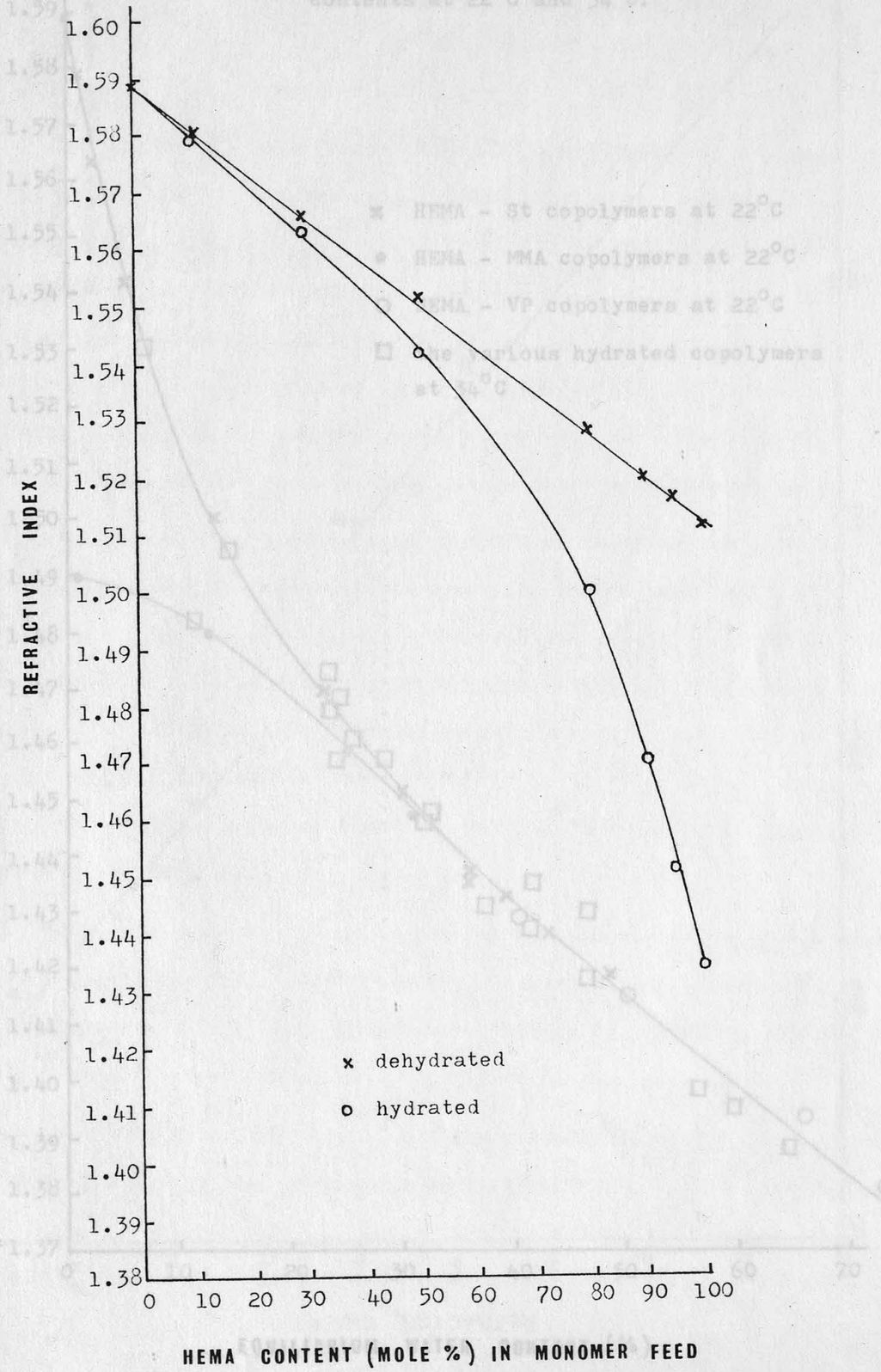


Figure II.13 Refractive indices of various hydrated copolymers with various equilibrium water contents at 22°C and 34°C.

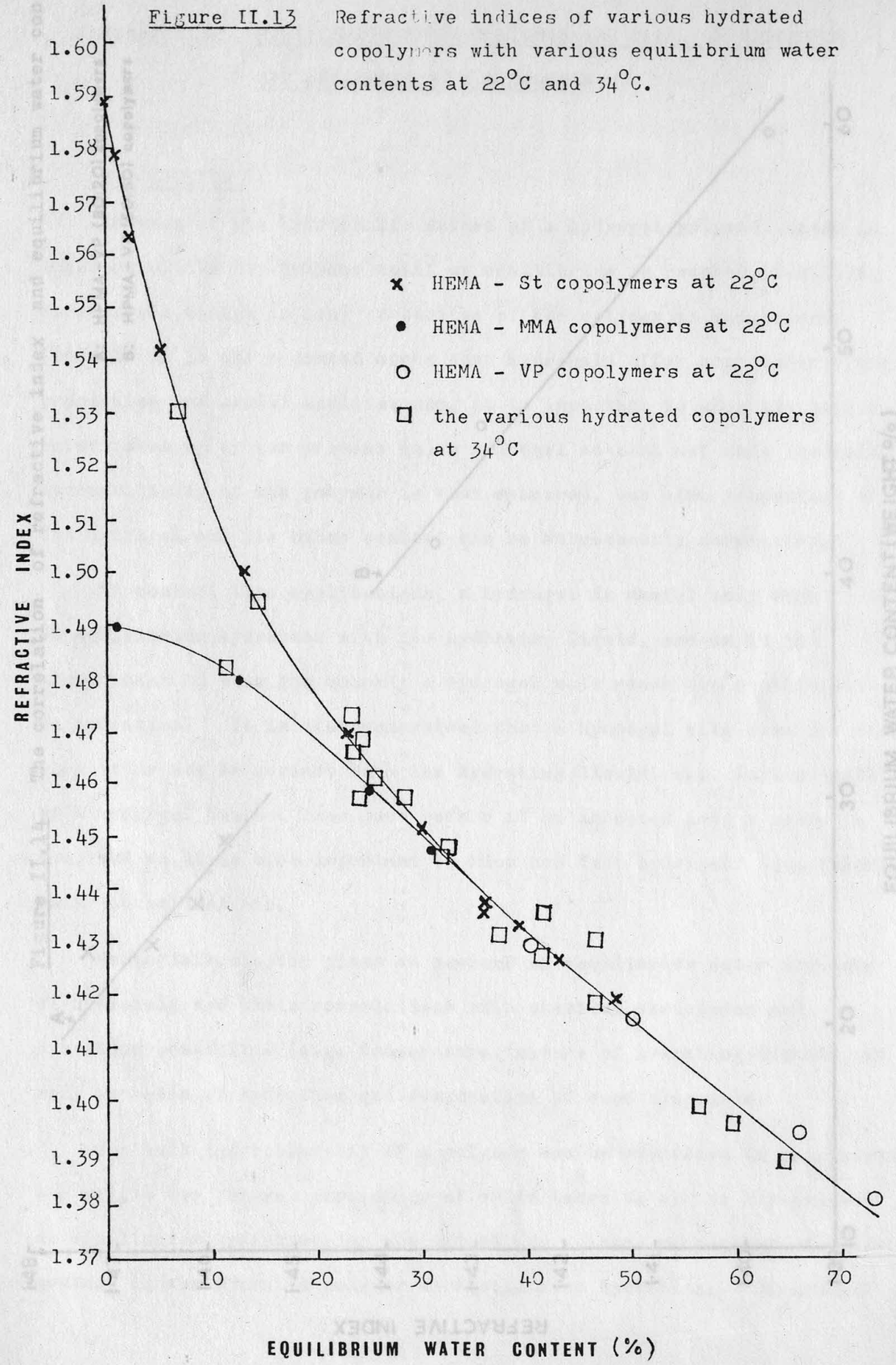
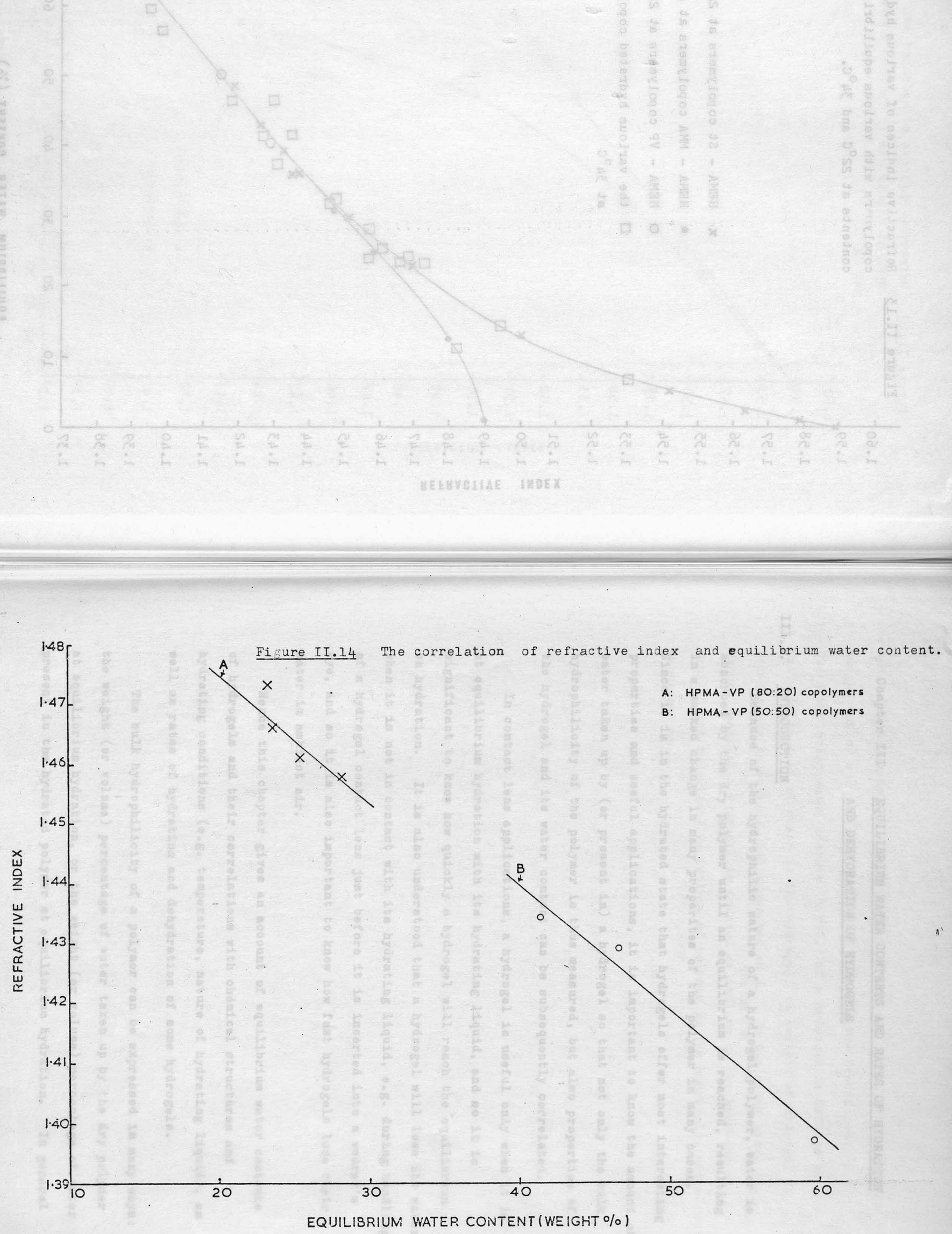


Figure II.14 The correlation of refractive index and equilibrium water content.



Chapter III. EQUILIBRIUM WATER CONTENTS AND RATES OF HYDRATION AND DEHYDRATION OF HYDROGELS

III.1. INTRODUCTION

Because of the hydrophilic nature of a hydrogel polymer, water is absorbed by the dry polymer until an equilibrium is reached, resulting in a marked change in many properites of the polymer in many cases. Since it is in the hydrated state that hydrogels offer most interesting properties and useful applications, it is important to know the amount of water taken up by (or present in) a hydrogel so that not only the bulk hydrophilicity of the polymer is thus measured, but also properties of the hydrogel and its water content can be subsequently correlated.

In contact lens applications, a hydrogel is useful only when it is at equilibrium hydration with its hydrating liquid, and so it is significant to know how quickly a hydrogel will reach the equilibrium on hydration. It is also understood that a hydrogel will lose its water when it is not in contact with its hydrating liquid, e.g. during handling of a hydrogel contact lens just before it is inserted into a wearer's eye, and so it is also important to know how fast hydrogels lose their water in ambient air.

Hence this chapter gives an account of equilibrium water contents of hydrogels and their correlations with chemical structures and hydrating conditions (e.g. temperature, nature of hydrating liquid), as well as rates of hydration and dehydration of some hydrogels.

The bulk hydrophilicity of a polymer can be expressed in many ways: the weight (or volume) percentage of water taken up by the dry polymer at equilibrium hydration, or the weight (or volume) percentage of water present in the hydrated polymer at equilibrium hydration. In general

and in this thesis, an equilibrium water content of a hydrogel (on the weight basis) and a volume swell of a hydrogel are considered.

The equilibrium water content (E.W.C.) of a hydrogel is defined as the ratio of (weight of water in the hydrogel) to (weight of hydrogel at equilibrium hydration), and expressed as %. On the other hand, the volume swell of a hydrogel is defined as the ratio of (volume of the hydrogel at equilibrium hydration) to (volume of polymer in the hydrogel), and also expressed as %.

III.2. EXPERIMENTAL

III.2.A. Equilibrium Water Content (E.W.C.).

Most of the copolymer and terpolymer sheets made and mentioned in Chapter II were used in the studies of E.W.C. The E.W.C. of a hydrated sheet at room temperature was measured as follows:

Several disc samples were cut with a cork-borer (about 1 cm. in diameter) from each hydrated polymer sheet which had been allowed to hydrate in distilled water at room temperature (22°C) for at least two weeks, by which the samples were thought to reach equilibrium hydration under that experimental condition. The surface water of each hydrated sample was removed with a soft tissue paper and the sample was quickly transferred to and weighed in a closed weighing bottle. The hydrated sample was then dried in air and then in a vacuum oven at 60°C for about 8 hours when a constant weight of the dehydrated sample was obtained when weighed again. The E.W.C. (i.e. W_W/W_G) of the hydrogel was then calculated using Equation III.1.

$$\frac{W_W}{W_G} = \frac{\text{wt. of hydrated sample - wt. of dehydrated sample}}{\text{wt. of hydrated sample}} \times 100\%$$

$$= \frac{(W_G - W_P)}{W_G} \times 100\% \qquad \text{(III.1)}$$

The E.W.Cs of two or three samples were determined in this way for each material, and an average E.W.C. was taken.

To study the effect of hydration temperature on the E.W.C. of a hydrated copolymer, two samples were equilibrated in distilled water at various temperatures, 22°C, 34°, 40°, 50°, 60° and 70°C (in ascending order) each for one week, and the E.W.Cs were measured in the same way as mentioned before.

The effect of the concentration of saline (NaCl) solution hydrating a hydrogel material was studied by equilibrating dehydrated samples of some of the materials [PHEMA, HEMA-St (90:10) copolymer and HPA-VP-St (45:45:10) terpolymer] in NaCl solutions of various concentrations (0.9, 1.5 and 2.0% w/v) for two weeks and then measuring their E.W.Cs as above.

III.2.B. Rates of Hydration and Dehydration

To study the rate of hydration of a xerogel (dehydrated sample), the hydrogel disc was first allowed to dehydrate slowly in air and then in a vacuum oven at 50°C, ensuring a reasonably flat dehydrated disc (xerogel) was obtained. The dehydrated sample was weighed in a closed weighing bottle, and then transferred to a beaker of distilled water to hydrate, when a stop-watch was also started. At time intervals, the hydrating sample was removed from the water and immediately its surface water was removed with a tissue paper and the sample was weighed again in a closed weighing bottle, and then it was quickly returned to water for further hydration. This procedure was continued until a constant weight of the hydrogel was obtained.

For dehydration study, the same hydrogel sample was left in water for one week to ensure complete equilibrium hydration, and after removal of surface water it was weighed on a small watch-glass covered with a crucible lid. The lid was then removed and the sample was allowed to dehydrate in ambient air. At time intervals, the lid was replaced on the watch-glass and the sample was weighed again. This procedure was continued until an approximately constant weight of the sample was obtained. The sample was then dried in a vacuum oven at 50°C for five hours until a final constant weight of the xerogel was obtained.

Three materials were used in the hydration/dehydration studies, namely PHEMA, HEMA-St (90:10) and HEMA-VP (50:50) copolymers.

III.3. RESULTS AND DISCUSSIONS ON EQUILIBRIUM WATER CONTENTS (E.W.C.)

The results on the measurements of E.W.Cs of the hydrated sheets are shown in Tables III.1 to III.4. (See Appendix 3).

In the determination of equilibrium water contents using the "weighing" method, the major contributions to inaccuracy are (a) incomplete removal of the surface water on the hydrated sample in question, or loss of water in the hydrogel due to slight pressure during removal of surface water, (b) incomplete drying of the hydrated sample before the weight of the dehydrated sample was taken, But the sample drying conditions of 65°C under vacuum for 8 hours used in this work seems to be satisfactory because further drying of the samples for another 9 hours did not alter the E.W.Cs, as shown in Table III.2 (see Appendix 3), and (c) the thickness of the sample seemed to affect the E.W.C. of a hydrogel determined, as shown on PHEMA hydrogel in Table III.2. The samples (lc_1, lc_2 and lc_3) were made under the same experimental conditions except at different thicknesses (0.2 to 0.7 mm), and their E.W.Cs were found to be less with a thicker sample, possibly because a thicker sample took a longer time to reach the equilibrium hydration. However the E.W.Cs of all the other hydrogels were compared at a sample thickness of about 0.2 mm (i.e. thickness of a normal contact lens.

All the E.W.C. results obtained here will show (a) how the chemical structures, copolymer compositions, cross-link densities, polymerisation conditions and hydration conditions (temperature, hydrating solution) affect E.W.C. and (b) possibility of calculating volume and linear swells of a hydrogel from its E.W.C.

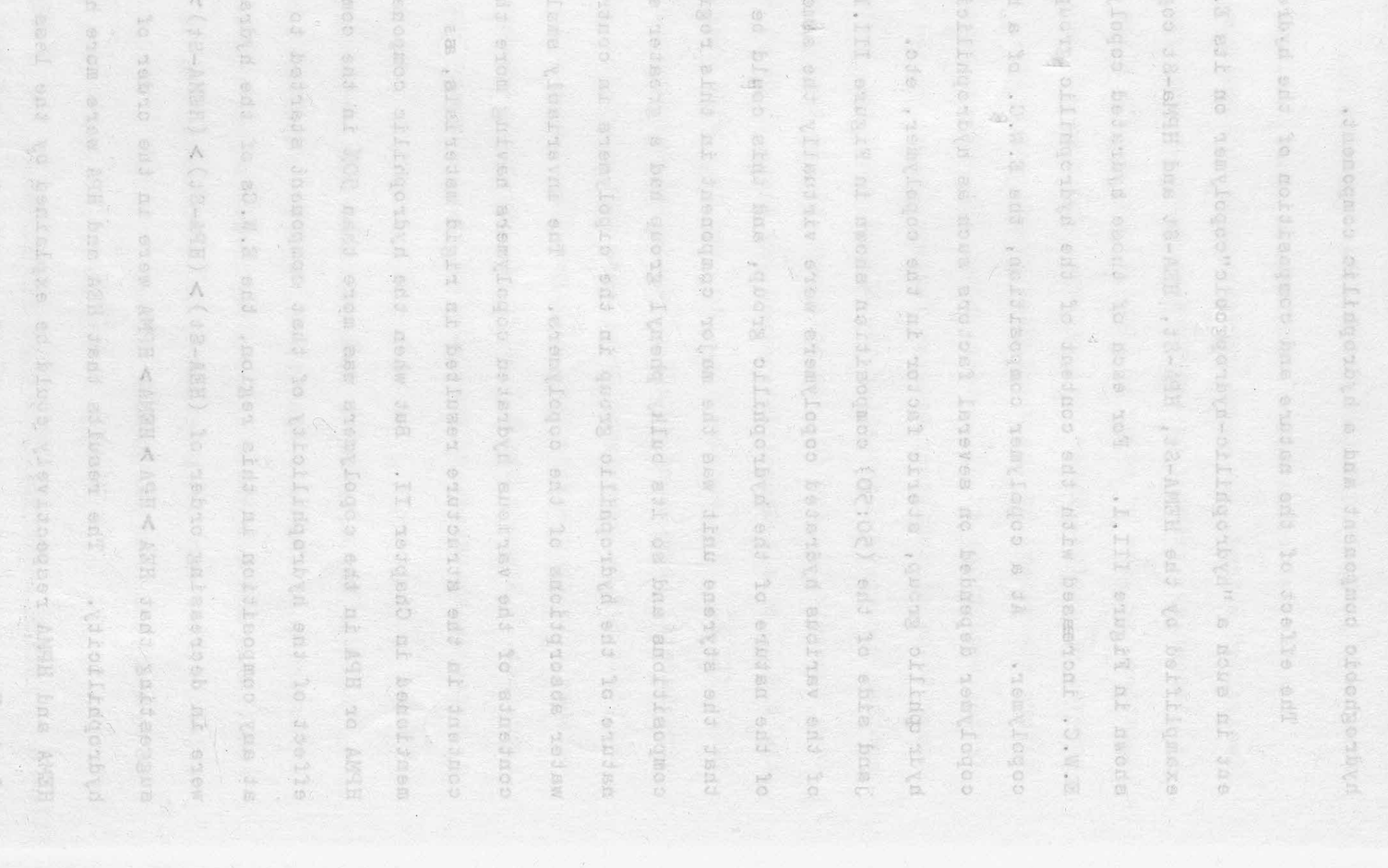
III.3.A. Factors Affecting E.W.Cs of Hydrogels.

III.3.A.1. Effect of Chemical Structures and Compositions

Most of the copolymers studied in this work are composed of a hydrophobic component and a hydrophilic component.

The effect of the nature and composition of the hydrophilic component in such a "hydrophilic-hydrophobic"copolymer on its E.W.C. was examplified by the HEMA-St, HPA-St, HEA-St and HPMA-St copolymers, as shown in Figure III.1. For each of those hydrated copolymers, the E.W.C. increased with the content of the hydrophilic group in the copolymer. At a copolymer composition, the E.W.C. of a hydrated copolymer depended on several factors such as hydrophilicity of the hydrophilic group, steric factor in the copolymer, etc. On the left-hand side of the (50:50) composition shown in Figure III.1, the E.W.Cs of the various hydrated copolymers were virtually the same, independent of the nature of the hydrophilic group, and this could be explained by that the styrene unit was the major component in this region of copolymer compositions and so its bulky phenyl group had a greater effect than the nature of the hydrophilic group in the copolymers in controlling the water absorptions of the copolymers. The invariably small water contents of the various hydrated copolymers having more than 50% styrene content in the structure resulted in rigid materials, as observed and mentioned in Chapter II. But when the hydrophilic component (HEMA,HEA, HPMA or HPA in the copolymers was more than 50% in the composition, the effect of the hydrophilicity of that component started to manifest, and at any composition in this region, the E.W.Cs of the hydrated copolymers were in decreasing order of (HEA-St) > (HPA-St) > (HEMA-St) > (HPMA-St), suggesting that HEA > HPA > HEMA > HPMA were in the order of decreasing hydrophilicity. The results that HEA and HPA were more hydrophilic than HEMA and HPMA respectively could be explained by the less steric hindrance of the H group on the main chain of the acrylate than the methyl group

Figure III.1 The equilibrium water contents of some styrene copolymers of various compositions at 22°C.



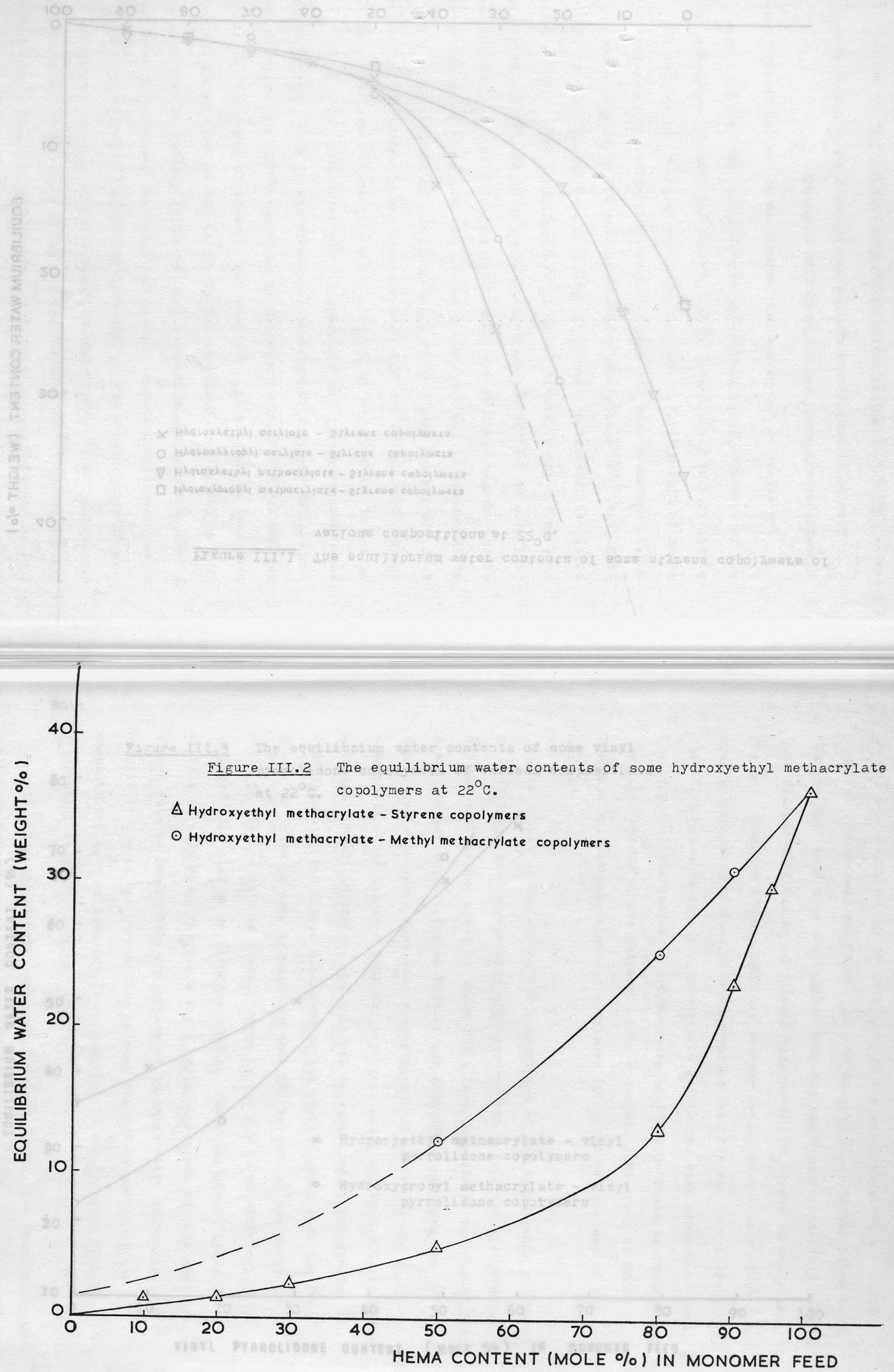
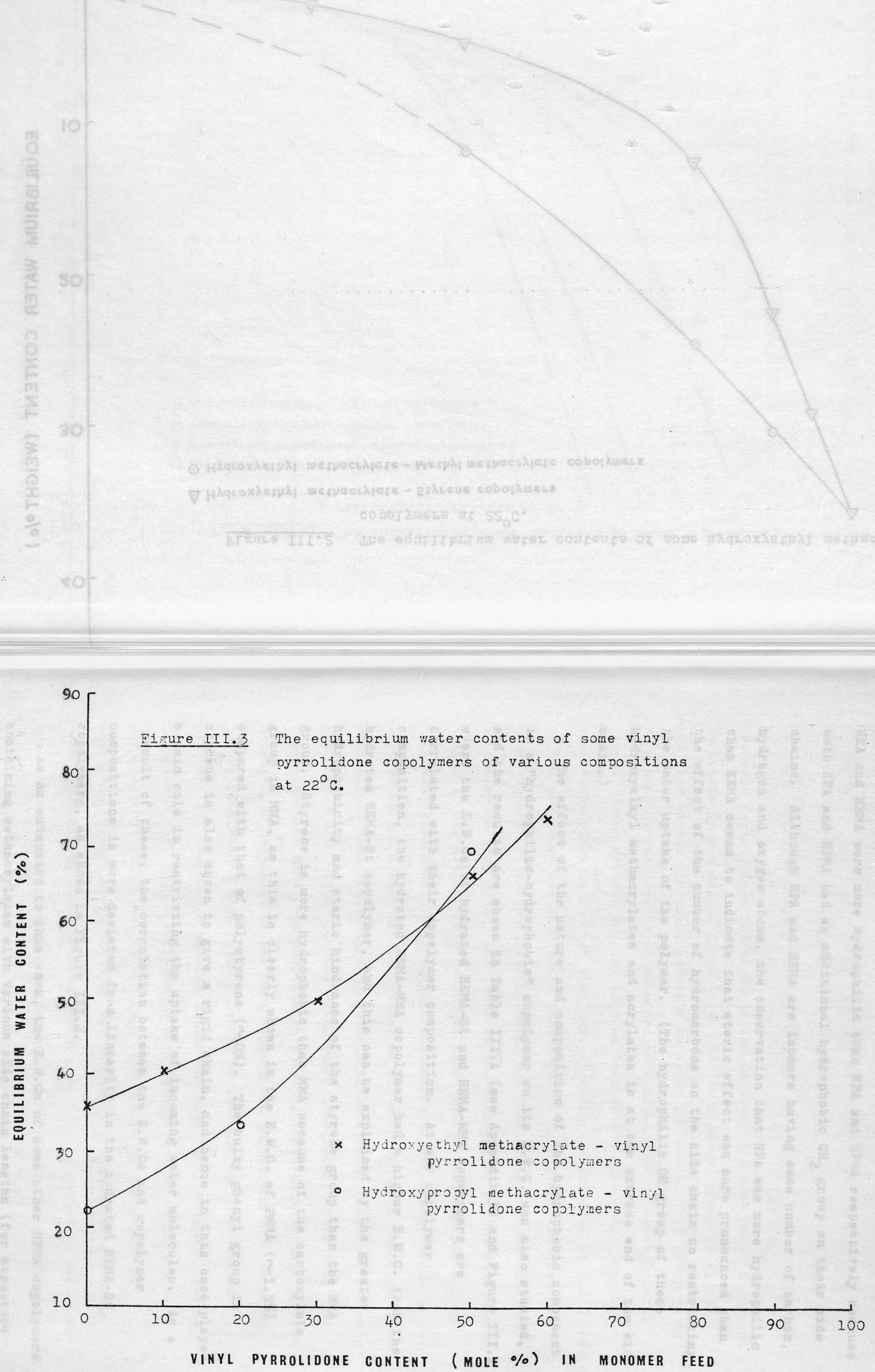


Figure III.2 The equilibrium water contents of some hydroxyethyl methacrylate copolymers at 22°C.

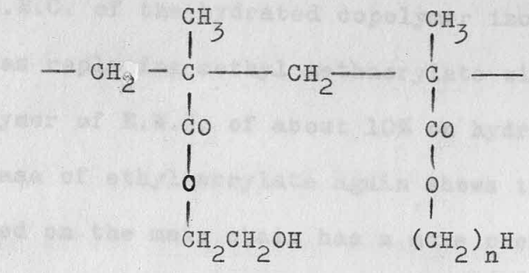
Figure III.3 The equilibrium water contents of some vinyl pyrrolidone copolymers of various compositions at 22°C.



on the main chain of the methacrylate on the water absorption. (For chemical structures, see Figure II.1 in Chapter II). On the other hand, HEA and HEMA were more hydrophilic than HPA and HPMA respectively because both HPA and HPMA had an additional hydrophobic CH_2 group on their side chains. Although HPA and HEMA are isomers having same number of carbon, hydrogen and oxygen atoms, the observation that HPA was more hydrophilic than HEMA seems to indicate that steric effect was more pronounced than the effect of the number of hydrocarbons on the side chain on restricting the water uptake of the polymer. (The hydrophilic OH group of these hydroxyalkyl methacrylates and acrylates is at the extreme end of the side chain.)

The effect of the nature and composition of the hydrophobic component in a "hydrophilic-hydrophobic" copolymer on its E.W.C. was also studied, and the results are shown in Table III.1 (see Appendix 3) and Figure III.2 where the E.W.Cs of hydrated HEMA-St and HEMA-MMA copolymers are correlated with their copolymer composition. At any copolymer composition, the hydrated HEMA-MMA copolymer had a higher E.W.C. than the hydrated HEMA-St copolymer, and this can be explained by the greater hydrophobicity and steric hindrance of the styrene group than the MMA group. Styrene is more hydrophobic than MMA because of the carboxylate group in MMA, as this is clearly shown in the E.W.C. of PMMA ($\sim$1.5%) compared with that of polystyrene ($\sim$0%). The bulky phenyl group in styrene is also known to give a rigid chain, and hence in this case played a main role in restricting the uptake of incoming water molecules. As a result of these, the correlation between the E.W.Cs and copolymer compositions is more deviated from linearity in the hydrated HEMA-St copolymers, as shown in Figure III.2.

As an extension to this work, the E.W.Cs of some other HEMA copolymers containing methacrylates with various side chain lengths (for structure see below) have been studied[206] at the 50:50 composition, and found to



decrease rapidly from 16% to 8% as the methacrylate side chain increased in length from a methyl group (i.e. n=1) to a propyl group (i.e. n=3), and then became virtually constant at about 7% from a butyl group (n=4) to a hexyl group (n=6) or a branched side chain such as 2-ethyl hexyl group. These results suggest that as long as the side chain length of the methacrylate is shorter than that of HEMA, the increase in the number of CH_2 groups on the side chain from a methyl to a propyl reduces the water uptake of the copolymer although the flexibility of the side chain increases in that order, and that, on the other hand, if the side chain length of the methacrylate is longer than that of HEMA (say from butyl onwards), the effect of increasing the number of CH_2 groups seems to be balanced by the effect of flexibility of the side chain on the water absorption. The result that hydrated HEMA-St (50:50) copolymer had only about 5% E.W.C. (found in this work), which is slightly lower than the other hydrated HEMA copolymer having butyl or hexyl methacrylate as the hydrophobic component at the same composition (E.W.C.7%)[206] might be due to the proximity of the rigid Ph group to the main chain in styrene, and/or slightly different experimental conditions used in both works.[206] The author also found that (i) by replacing a butyl group ($-CH_2CH_2CH_2CH_3$) with a 2-ethoxyethyl group ($-CH_2CH_2OCH_2CH_3$), the E.W.C. of the copolymer increased from about 7% to about 14%, apparently because of the polar (and hence hydrophilic) ethereal oxygen and (ii) by replacing methyl methacrylate with its structural isomer ethyl acrylate in the copolymer,

the E.W.C. of the hydrated copolymer increased from about 16% to about 19% whereas replacing methyl methacrylate with ethyl methacrylate gave a copolymer of E.W.C. of about 10% on hydration. This reversing order in the case of ethyl acrylate again shows that the less steric H group located on the main chain has a more profound effect than the increase in side chain length on controlling the water absorption of the HEMA copolymer.

The E.W.Cs of copolymers having two hydrophilic components were also studied in this work, and Figure III.3 shows the E.W.Cs of HEMA-VP and HPMA-VP copolymer hydrogels vs. their copolymer compositions. The E.W.C. of the hydrogels increases with the VP content in the copolymers, illustrating that VP is more hydrophilic than HEMA and HPMA. In fact poly(vinyl pyrrolidone) is known to be soluble in water, and the greater hydrophilicity of VP could possibly be explained by the presence of polar amide group near the vinyl pyrrolidone backbone.

The E.W.Cs of some terpolymers containing two hydrophilic components and one hydrophobic component (e.g. styrene) were also determined and are shown in Table III.1 (see Appendix 3). The results show again that (a) HPA was more hydrophilic than HEMA (by comparing sheets 54 and 55, or 56 and 62), (b) VP was more hydrophilic than HEMA (by comparing sheets 53 and 55), and (C) VP was more hydrophilic than HPA (by comparing sheets 53 and 54, or 56 to 61). So all these results obtained for the terpolymers substantiate the conclusions made earlier for the copolymers.

However the equilibrium water content of a hydrated copolymer (or terpolymer) is dependent on its chemical structure (such as hydrophilicity, steric factor and flexibility of its components) and copolymer composition.

For ease of discussion, the terms "high water content hydrogels" and "low water content hydrogels" will be used hereafter in this thesis,

referring to hydrogels having E.W.Cs more than or less than about 40% (E.W.C. of PHEMA hydrogel)respectively.

II.3.A.2. Effect of Cross-link Density

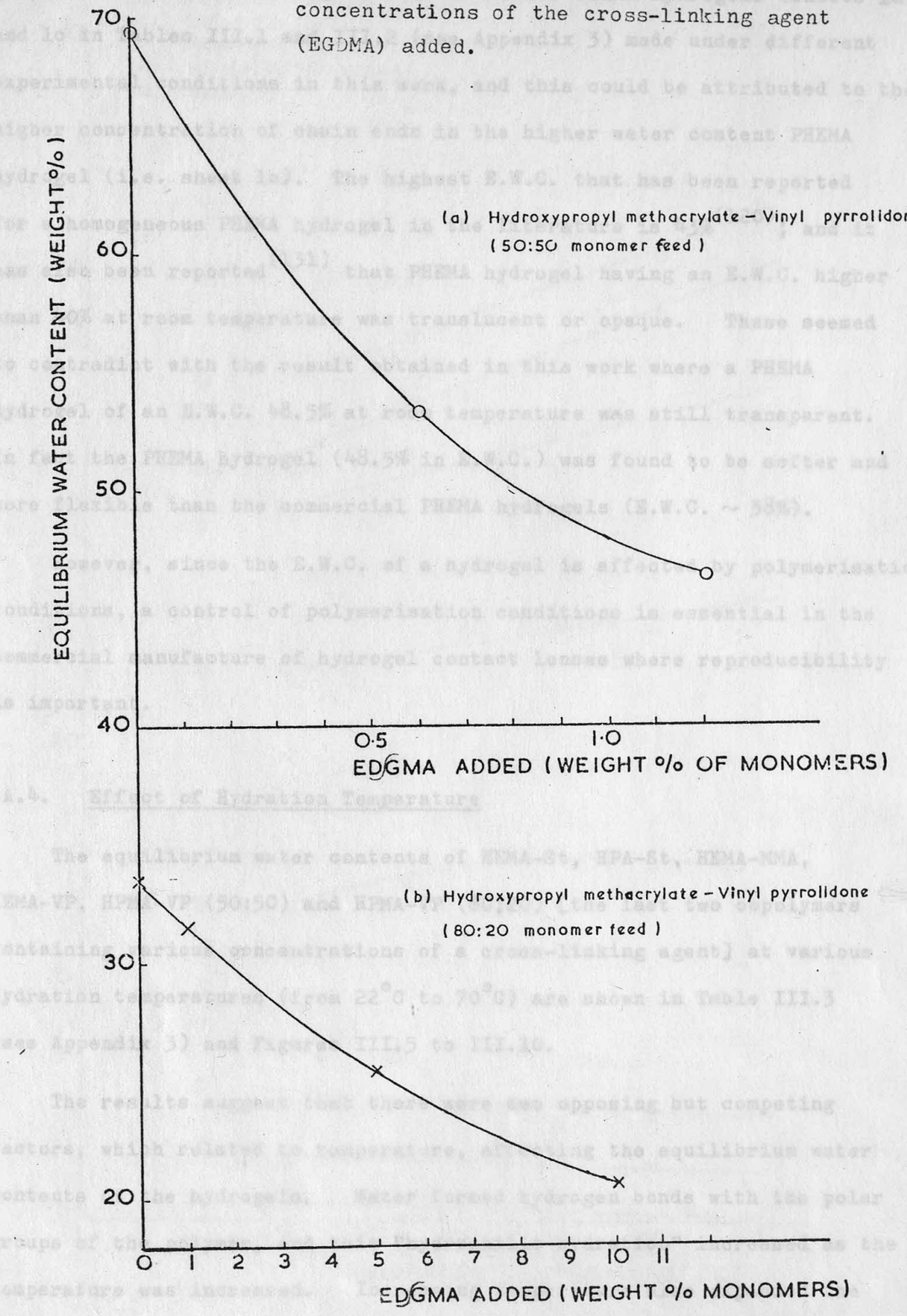
The effect of crosslink density of a hydrogel on its E.W.C. was studied by determining the E.W.Cs of various hydrogels of a copolymer made by incorporating various concentrations of a cross-linking agent (EGDMA) into its polymerisation mixture. The results which are shown in Table III.1 (see Appendix 3) and Figure III.4 for HPMA-VP (80:20) and HPMA-VP (50:50) hydrogels indicate that the E.W.C. of the high water content hydrogel [HPMA-VP (50:50)] is more dependent of its cross-link density than the low water content hydrogel [HPMA (80:20)]. It is very likely that water is a thermodynamically poorer solvent for HPMA-VP (80:20) copolymer than HPMA-VP (50:50) copolymer, and so a tighter network due to an increase in the concentration of the cross-linking agent did not affect the E.W.C. of HPMA-VP (80:20) hydrogel as much as in the case of HPMA-VP (50:50) hydrogel. It has been reported[128] that cross-link density (between 0.3% to 3.0% of tetraethylene glycol dimethacrylate) had little effect on the E.W.C. of PHEMA hydrogel, which is somewhere between the E.W.Cs of the two HPMA-VP copolymer hydrogels studied here.

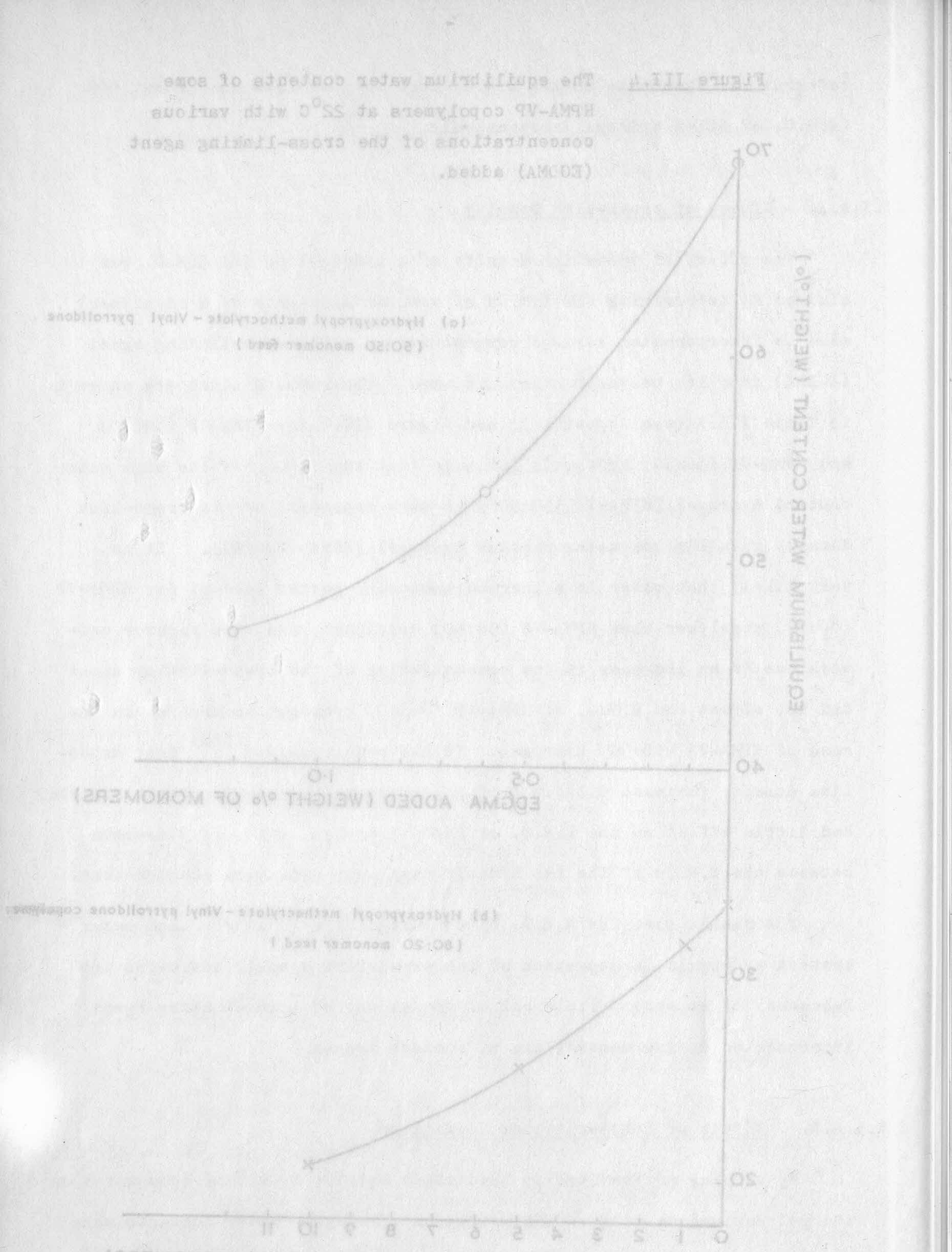
The result that the E.W.C. of a hydrogel (particularly high water content hydrogel) is dependent of its cross-link density indicates the necessity of an accurate control of the amount of cross-linking agent incorporated during manufacture of contact lenses.

III.3.A.3. Effect of Polymerisation Conditions

By varying polymerisation conditions such as initiator concentration and polymerisation time, PHEMA hydrogels of various water contents were obtained. The PHEMA hydrogel (sheet 1b) made with a higher initiator

Figure III.4 The equilibrium water contents of some HPMA-VP copolymers at 22°C with various concentrations of the cross-linking agent (EGDMA) added.





concentration and a shorter polymerisation time was found to have an E.W.C.of 48.5% which is higher than the other PHEMA hydrogels (sheets 1a and 1c in Tables III.1 and III.2 (see Appendix 3) made under different experimental conditions in this work, and this could be attributed to the higher concentration of chain ends in the higher water content PHEMA hydrogel (i.e. sheet 1b). The highest E.W.C. that has been reported for a homogeneous PHEMA hydrogel in the literature is 43%[120]; and it has also been reported[131] that PHEMA hydrogel having an E.W.C. higher than 40% at room temperature was translucent or opaque. These seemed to contradict with the result obtained in this work where a PHEMA hydrogel of an E.W.C. 48.5% at room temperature was still transparent. In fact the PHEMA hydrogel (48.5% in E.W.C.) was found to be softer and more flexible than the commercial PHEMA hydrogels (E.W.C. ~ 38%).

However, since the E.W.C. of a hydrogel is affected by polymerisation conditions, a control of polymerisation conditions is essential in the commercial manufacture of hydrogel contact lenses where reproducibility is important.

III.3.A.4. Effect of Hydration Temperature

The equilibrium water contents of HEMA-St, HPA-St, HEMA-MMA, HEMA-VP, HPMA-VP (50:50) and HPMA-VP (80:20) [the last two copolymers containing various concentrations of a cross-linking agent] at various hydration temperatures (from 22°C to 70°C) are shown in Table III.3 (see Appendix 3) and Figures III.5 to III.10.

The results suggest that there were two opposing but competing factors, which related to temperature, affecting the equilibrium water contents of the hydrogels. Water formed hydrogen bonds with the polar groups of the polymer, and this "hydrophilic hydration" increased as the temperature was increased. Increasing temperature also expanded the

Figure III.5 The equilibrium water contents of HEMA-St copolymers at various temps.

A: HEMA-ST (10:90)
B: HEMA-ST (30:70)
C: HEMA-ST (50:50)
D: HEMA-ST (80:20)
E: HEMA-ST (90:10)
F: HEMA-ST (95:5)
H: HEMA+(0.9%) EGDMA
G: PHEMA

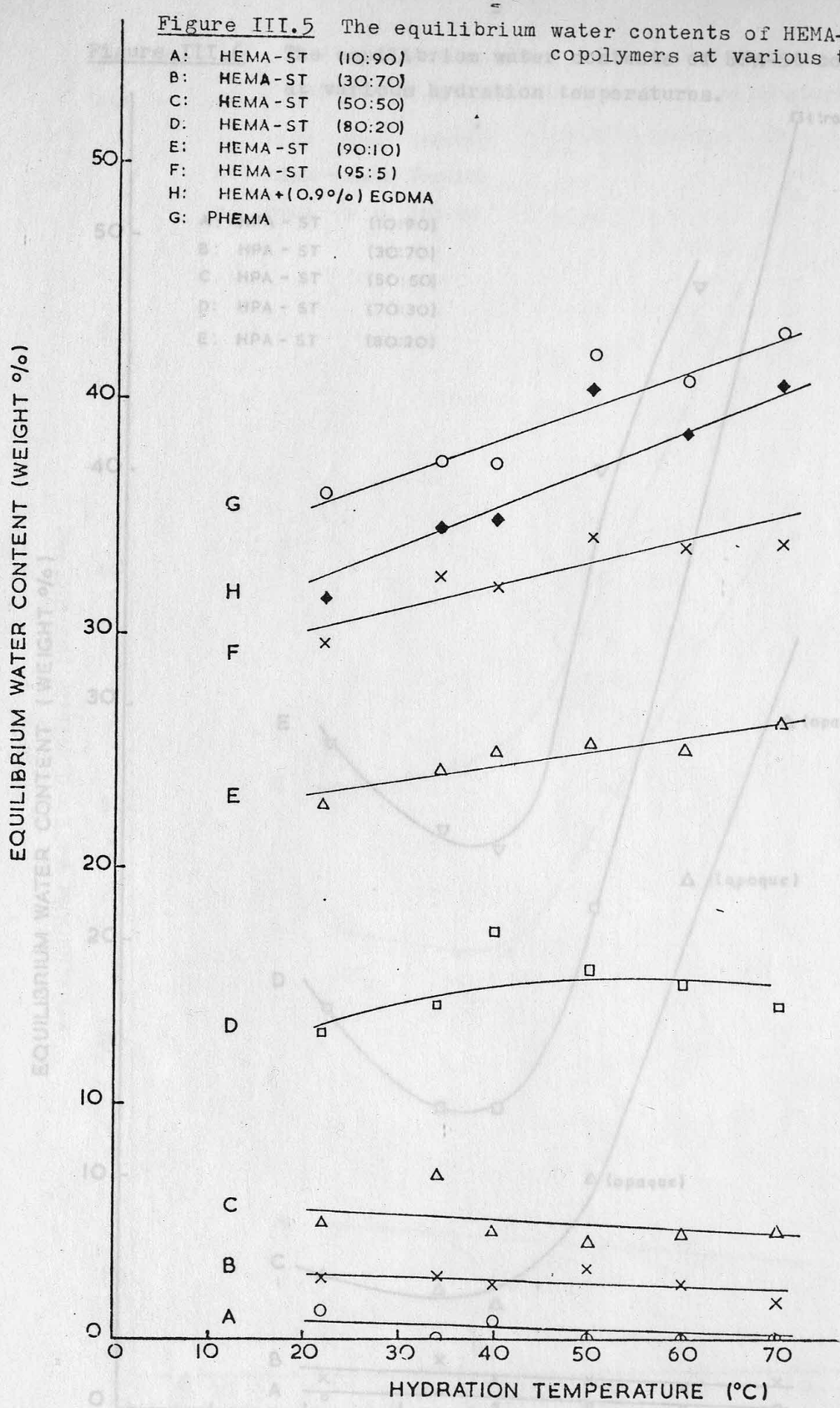


Figure III.6 The equilibrium water contents of HPA-St copolymers at various hydration temperatures.

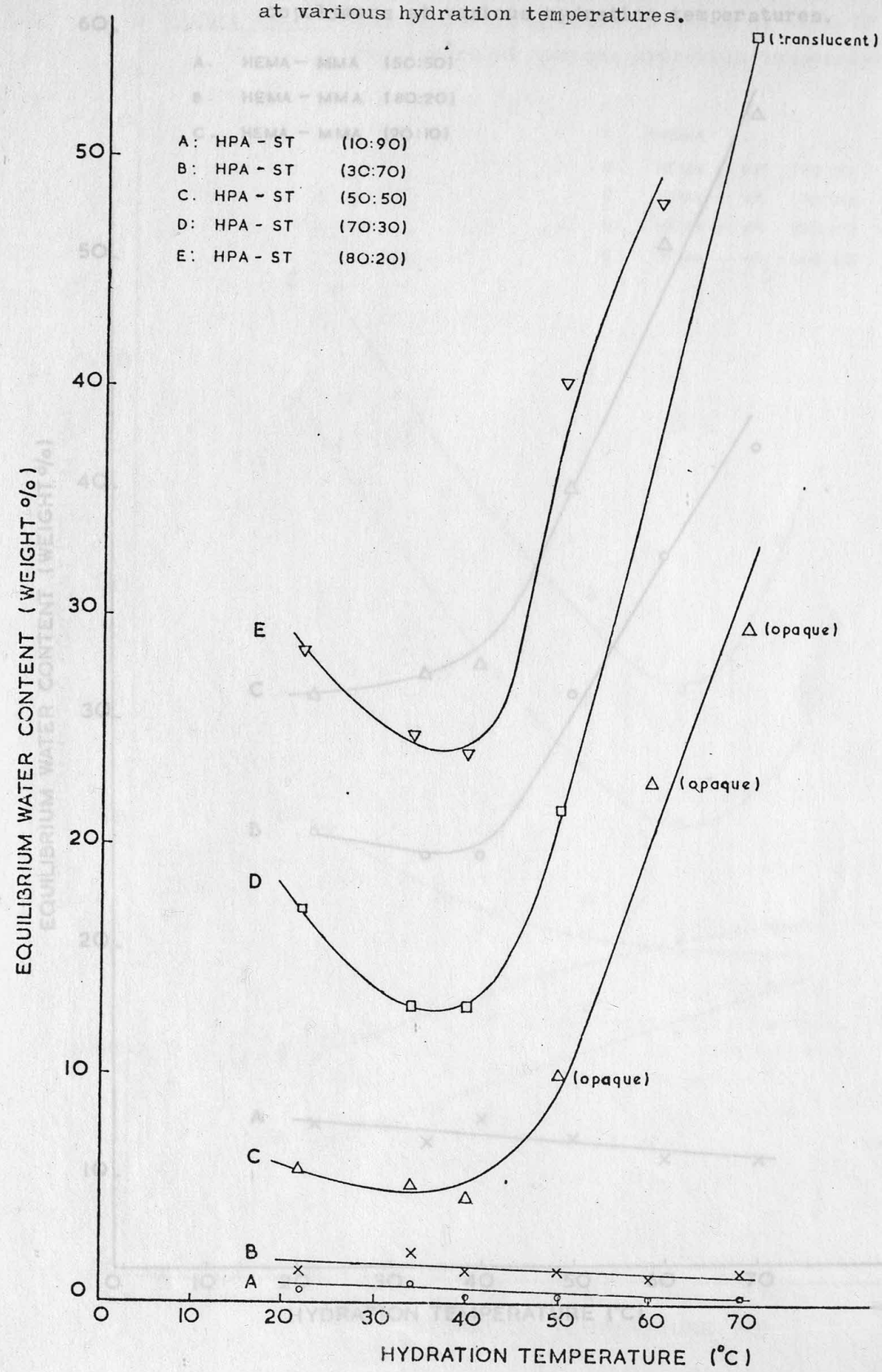


Figure III.7 The equilibrium water contents of HEMA-MMA copolymers at various hydration temperatures.

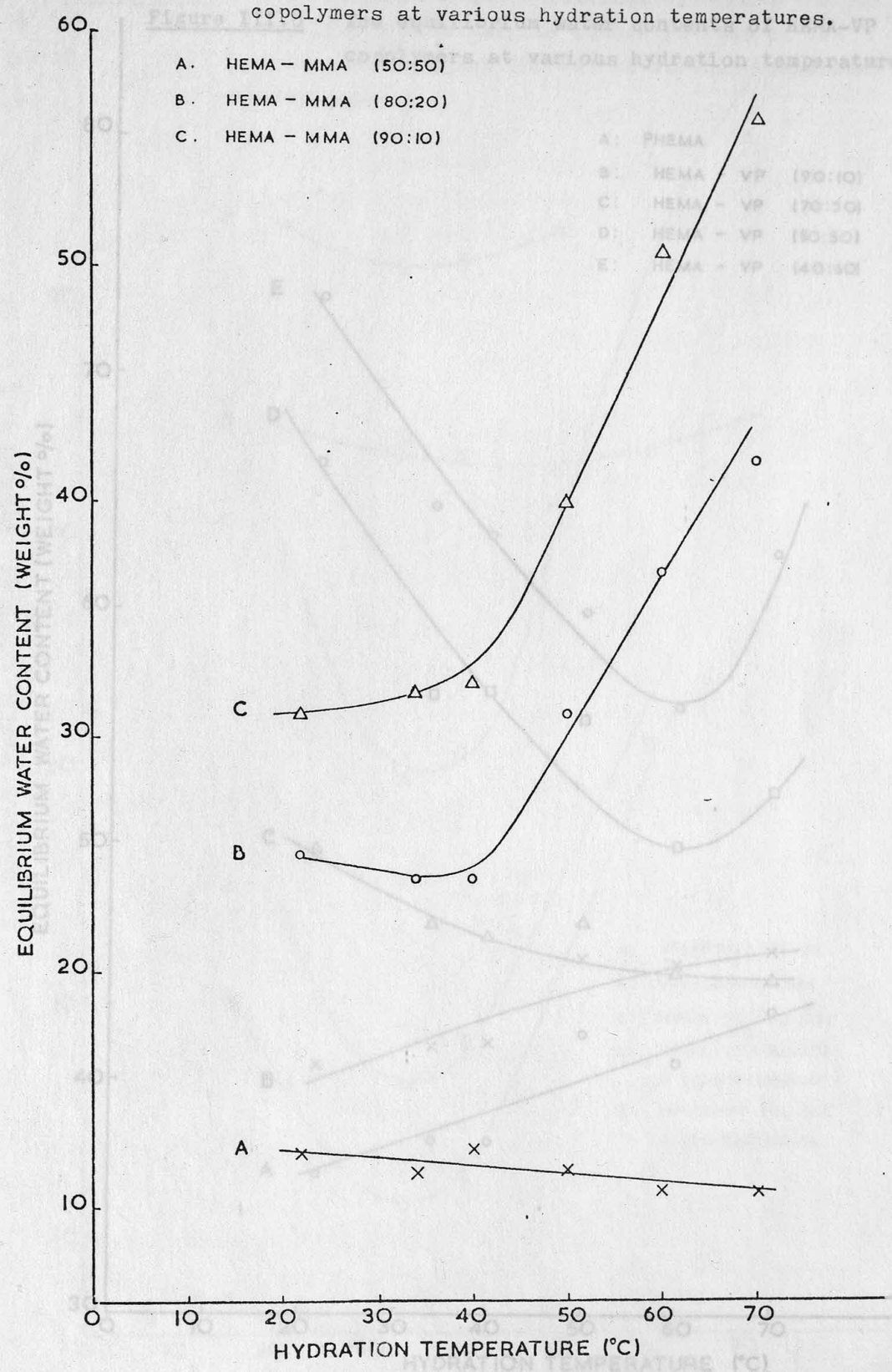


Figure III.8 The equilibrium water contents of HEMA-VP copolymers at various hydration temperatures.

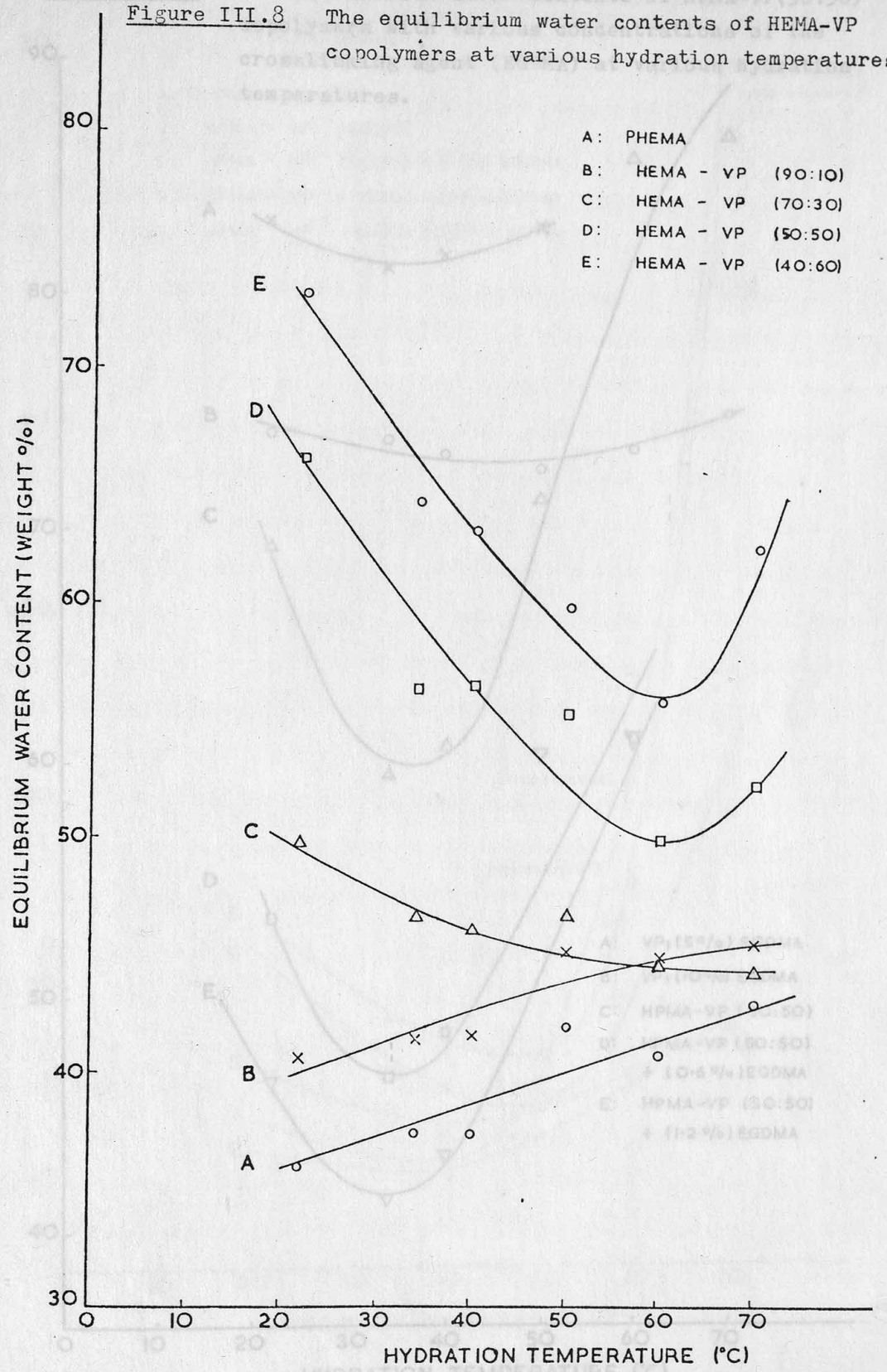


Figure III.9 The equilibrium water contents of HPMA-VP(50:50) copolymers with various concentrations of the crosslinking agent (EGDMA) at various hydration temperatures.

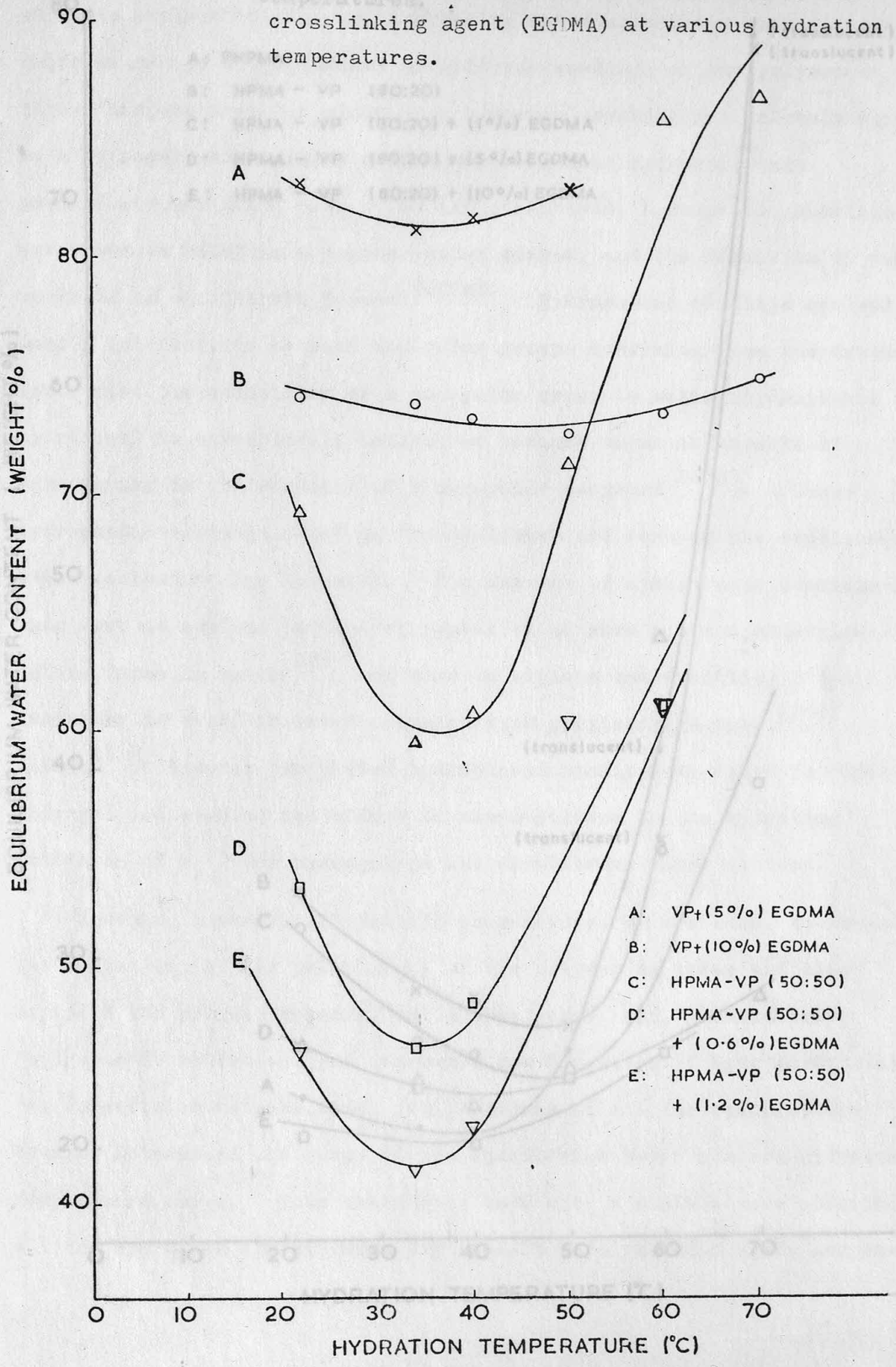
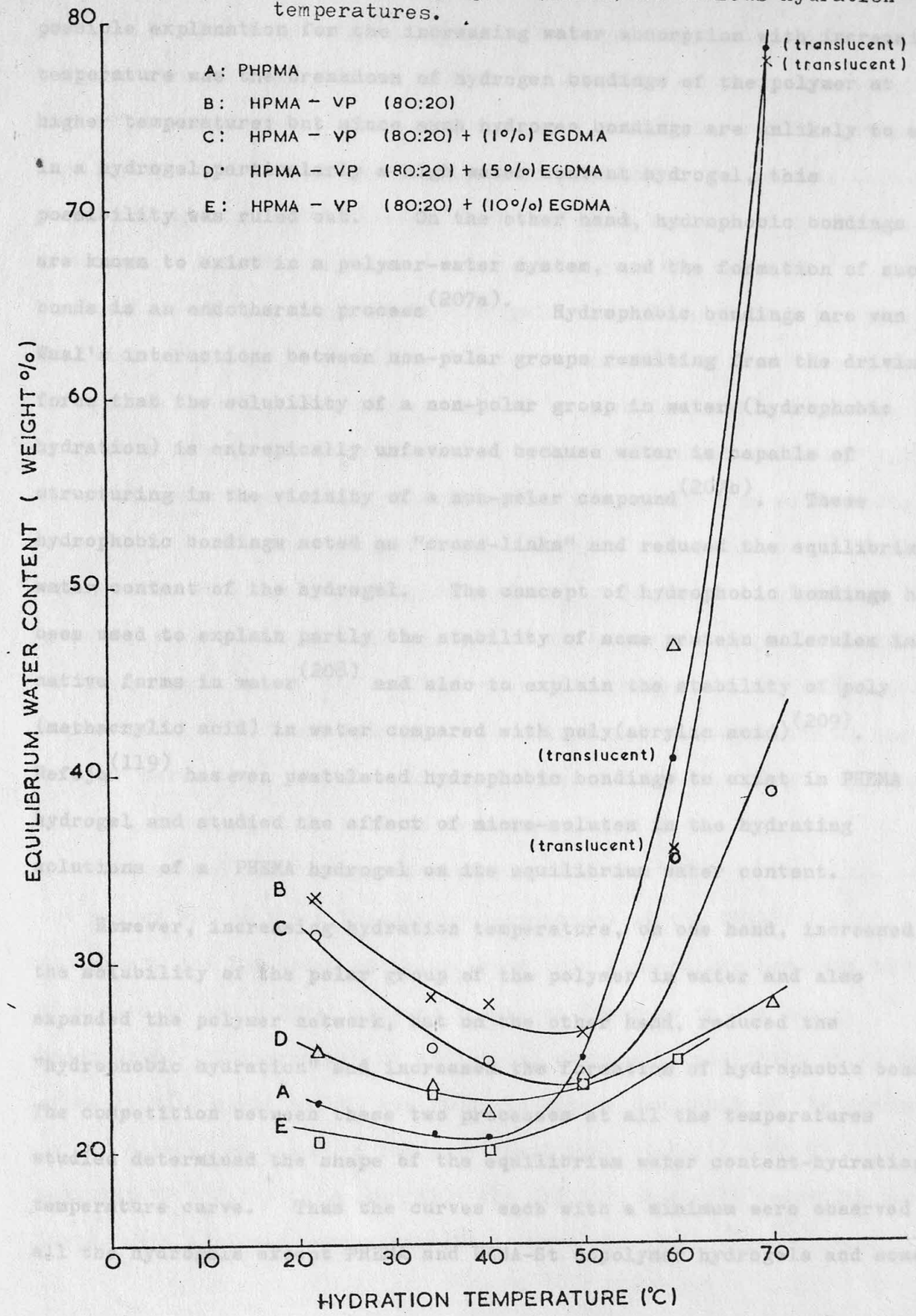
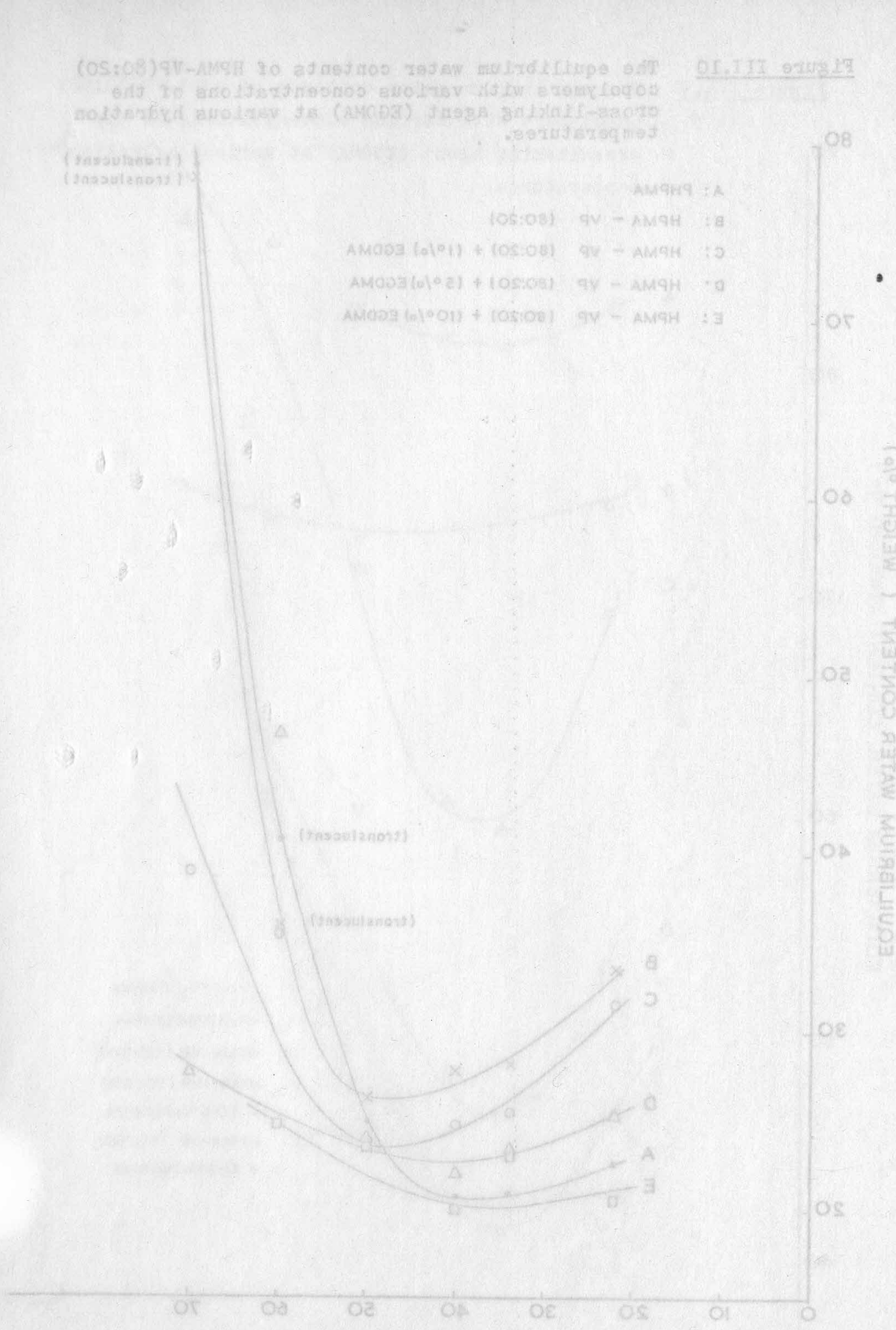


Figure III.10 The equilibrium water contents of HPMA-VP(80:20) copolymers with various concentrations of the cross-linking agent (EGDMA) at various hydration temperatures.





polymer network resulting in an increase in water absorption. Another possible explanation for the increasing water absorption with increasing temperature was the breakdown of hydrogen bondings of the polymer at higher temperature; but since such hydrogen bondings are unlikely to exist in a hydrogel,particularly a high water content hydrogel, this possibility was ruled out. On the other hand, hydrophobic bondings are known to exist in a polymer-water system, and the formation of such bonds is an endothermic process[207a]. Hydrophobic bondings are van der Waal's interactions between non-polar groups resulting from the driving force that the solubility of a non-polar group in water (hydrophobic hydration) is entropically unfavoured because water is capable of structuring in the vicinity of a non-polar compound[207b]. These hydrophobic bondings acted as "cross-links" and reduced the equilibrium water content of the hydrogel. The concept of hydrophobic bondings has been used to explain partly the stability of some protein molecules in their native forms in water[208] and also to explain the stability of poly (methacrylic acid) in water compared with poly(acrylic acid)[209]. Refojo[119] has even postulated hydrophobic bondings to exist in PHEMA hydrogel and studied the effect of micro-solutes in the hydrating solutions of a PHEMA hydrogel on its equilibrium water content.

However, increasing hydration temperature, on one hand, increased the solubility of the polar group of the polymer in water and also expanded the polymer network, but on the other hand, reduced the "hydrophobic hydration" and increased the formation of hydrophobic bonds. The competition between these two processes at all the temperatures studied determined the shape of the equilibrium water content-hydration temperature curve. Thus the curves each with a minimum were observed for all the hydrogels except PHEMA and HEMA-St copolymer hydrogels and some

hydrated copolymers in the non-gel state (see Figures III.5 to III.10). The equilibrium water content of PHEMA hydrogel has been reported (123,210a) to pass through a minimum at about 60°C, and this is inconsistent with the result obtained here. The exact reason is not clear, but one possible explanation is that different polymerisation conditions used in both works might end up with PHEMA hydrogel with different degrees of hydrophobic bondings. The slow increase in the curves over the range of temperature studied for HEMA-St hydrogels could be explained by the rigidity of the polymer chains, due to the presence of the bulky styrene groups, which restricted their rotation to give favourable positions for much more hydrophobic bond formation. For the low-water-content hydrated materials which were in the non-gel state, the equilibrium water contents were found to decrease slightly with hydration temperatures. Hydrophobic bondings seemed to be predominant in these copolymers (possibly due to a higher proportion of non-polar groups) and the rigidity of the polymer chains (below glass transition temperatures) restricted the formation of additional hydrophobic bondings and the expansion of the polymer networks. One exception was found for HPA-St (50:50) copolymer whose equilibrium water content decreased slightly from the room temperature and then increased abruptly at about 45°C. This could be the region of the glass transition temperature of the HPA-St (50:50) copolymer, and the chain rotation above this temperature resulted in an expansion of the polymer network and so an increase in its water absorption. The presence of additional chemical cross-links (i.e. more cross-linking agent added) was found to reduce the depth of the minimum on the curves, as shown in the poly(vinyl pyrrolidone), HPMA-VP (50:50) and HPMA-VP (80;20) (Figures III.9 and III.10), and such reduction was larger with a higher concentration of the cross-linking agent added.

This could be explained by that tighter polymer networks (i.e. more chemical cross-links) restricted the rotation of the polymer chains for maximum interaction of the non-polar groups (i.e. reduced formation of hydrophobic bondings) and also restricted the expansion of the polymer networks for water absorption, as the hydration temperature was increased.

At higher temperatures, some of the hydrogels were observed to become translucent [e.g. HPA-St (70:30), PHPMA, HPMA-VP (80:20)] or opaque [e.g. HPA-St (50:50)] (see Figures III.6 and III.10). This is thought to be due to the formation of a dispersion of separated water clusters and hydrogel phases (microsyneresis), and the slow relaxation of the polymer chains in expelling the micro-separated water out of the swollen network. Similar temperature-induced phase separation has also been observed[132] in PHEMA network swollen with n-butanol when the temperature is reduced. It has been concluded[132,210b] that the disperse system is in a non-equilibrium state, and complete disappearance of a temperature-induced turbidity might take months and is accompanied by a decrease in volume.

III.3.A.5. Effect of Concentration of Sodium Chloride Solutions

The effect of the concentration of sodium chloride in the hydrating solution on the equilibrium water contents of HPA-VP-St (45:45:10), PHEMA and HEMA-St (90:10) hydrogels are shown in Table III.4 (see Appendix 3) and Figure III.11.

The results show that the equilibrium water contents of the hydrogels decreased with increasing sodium chloride concentration, and that the effect was larger for the higher water content hydrogel i.e. HPA-VP-St (45:45:10). This could be explained by that the presence of sodium chloride in water enhanced the water structure, resulting in an increase in the hydrophobic bondings in the hydrogels and hence a decrease in the

equilibrium water contents of the hydrogels. The increase in the hydrophobic bondings in the HPA-VP-St (45:45:10) hydrogel was apparently larger than in the other two hydrogels. That PHEMA hydrogel deswells in sodium chloride solutions has also been reported previously[111,119].

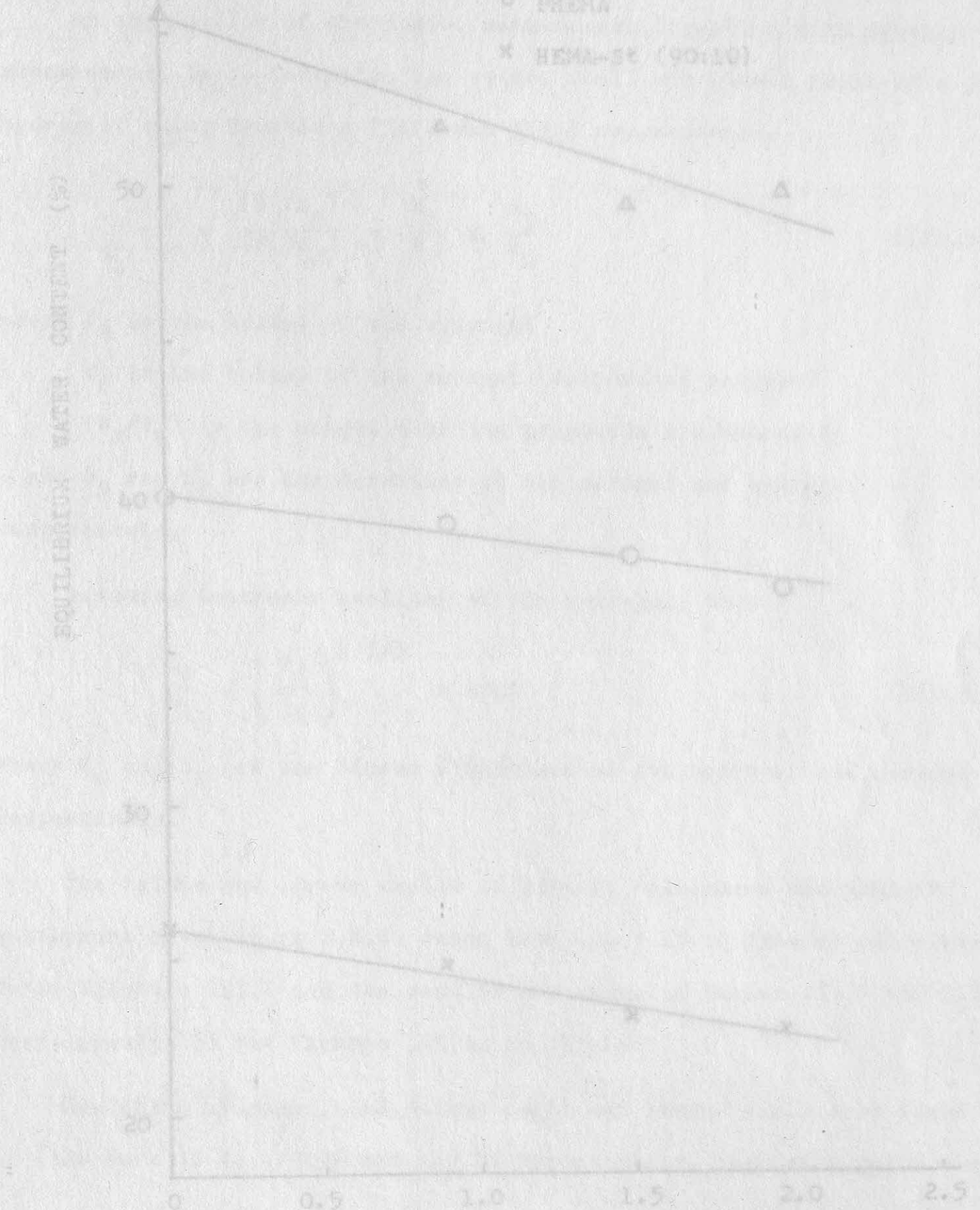
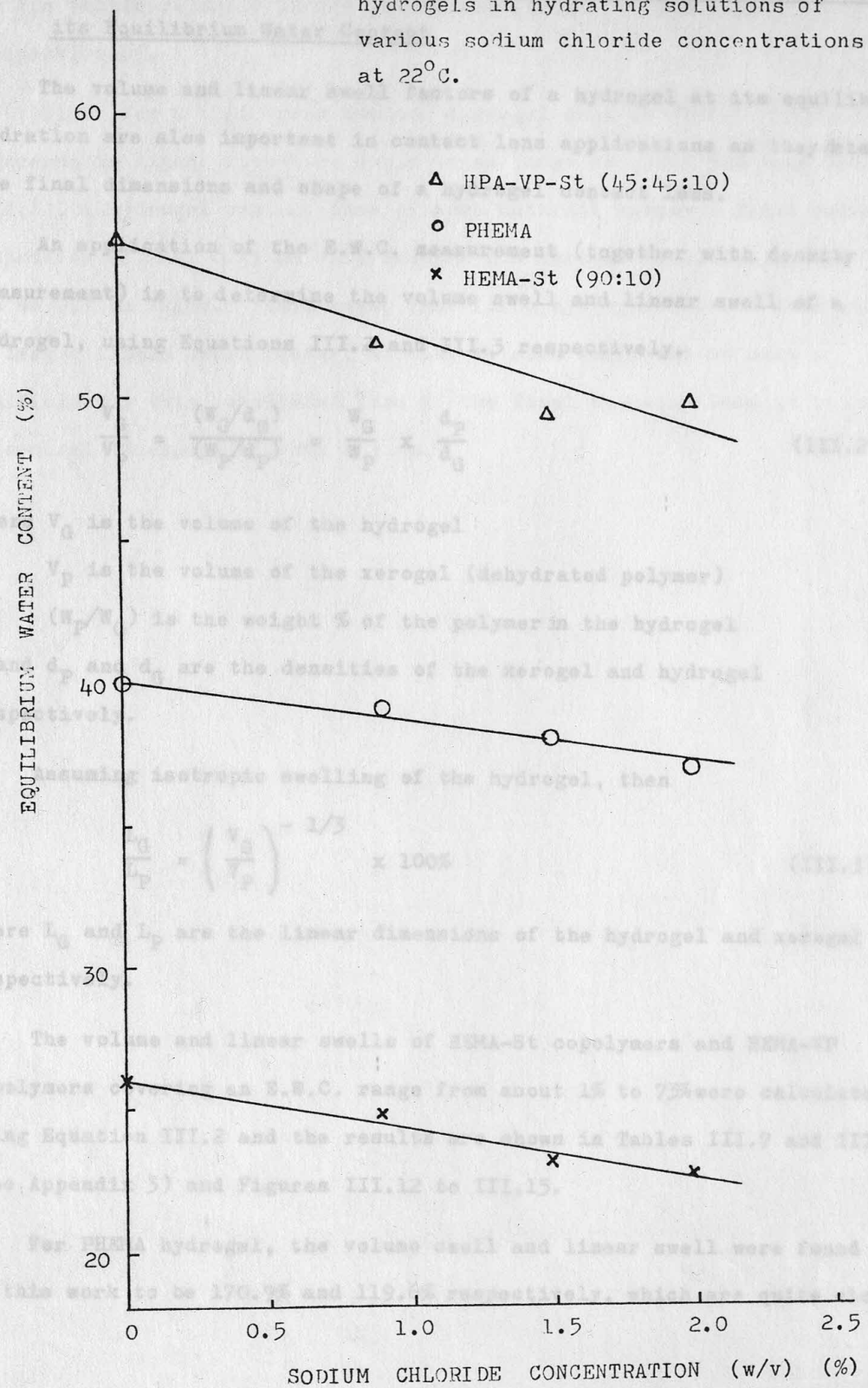
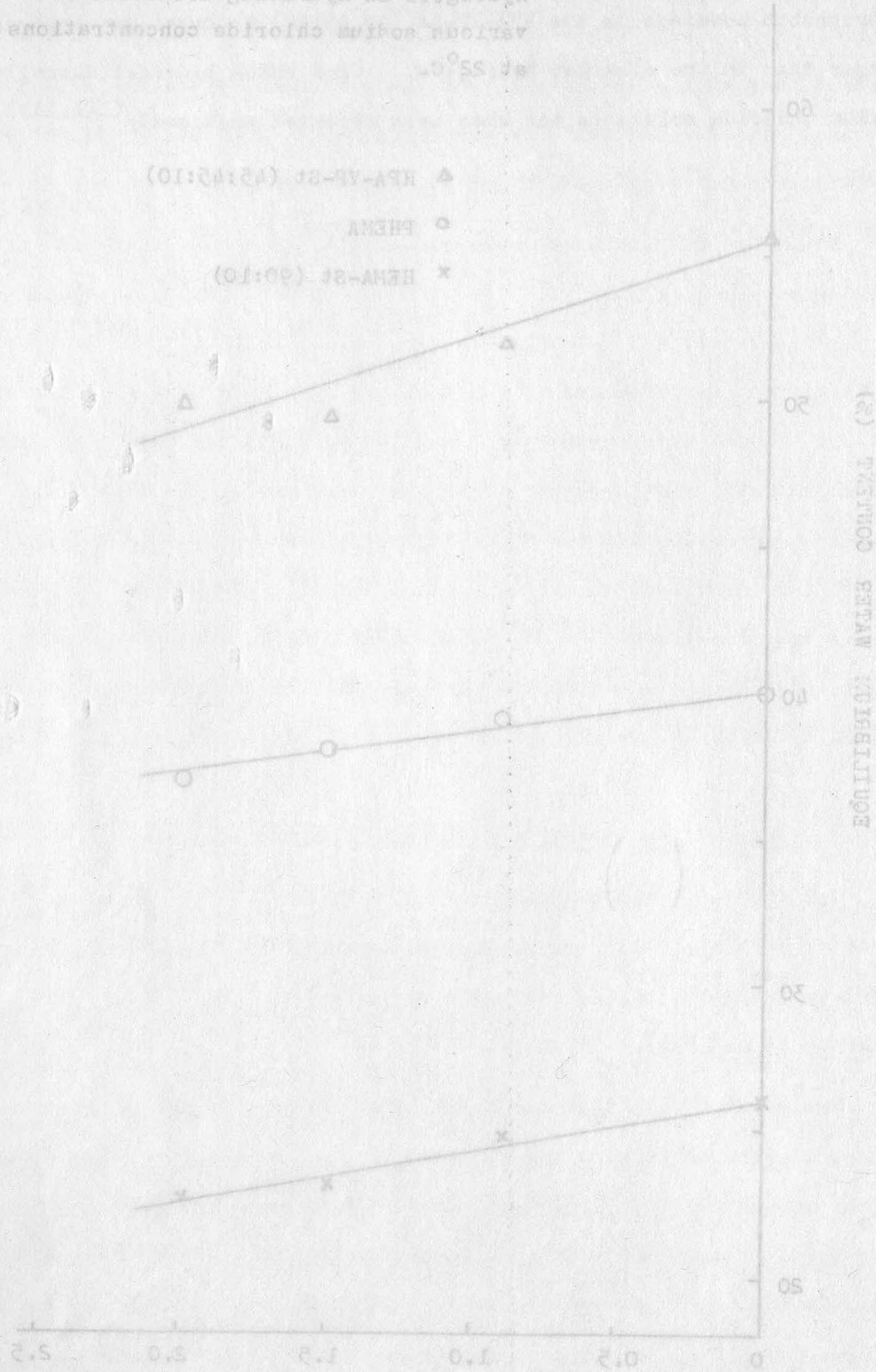


Figure III.11 The equilibrium water contents of some hydrogels in hydrating solutions of various sodium chloride concentrations at 22°C.





III.3.B. Determination of Volume Swell and Linear Swell of a Hydrogel from its Equilibrium Water Content

The volume and linear swell factors of a hydrogel at its equilibrium hydration are also important in contact lens applications as they deter the final dimensions and shape of a hydrogel contact lens.

An application of the E.W.C. measurement (together with density measurement) is to determine the volume swell and linear swell of a hydrogel, using Equations III.2 and III.3 respectively.

$$\frac{V_G}{V_P} = \frac{(W_G/d_G)}{(W_P/d_P)} = \frac{W_G}{W_P} \times \frac{d_P}{d_G} \qquad \text{(III.2)}$$

where V_G is the volume of the hydrogel

V_P is the volume of the xerogel (dehydrated polymer)

(W_P/W_G) is the weight % of the polymer in the hydrogel

and d_P and d_G are the densities of the xerogel and hydrogel respectively.

Assuming isotropic swelling of the hydrogel, then

$$\frac{L_G}{L_P} = \left(\frac{V_G}{V_P}\right)^{-1/3} \times 100\% \qquad \text{(III.3)}$$

where L_G and L_P are the linear dimensions of the hydrogel and xerogel respectively.

The volume and linear swells of HEMA-St copolymers and HEMA-VP copolymers covering an E.W.C. range from about 1% to 73% were calculated using Equation III.2 and the results are shown in Tables III.7 and III.8 (see Appendix 3) and Figures III.12 to III.15.

For PHEMA hydrogel, the volume swell and linear swell were found in this work to be 170.9% and 119.6% respectively, which are quite close

to the values reported in the literature [168%[129] and 119%[204] respectively].

Since for a high water content hydrogel such as HEMA-VP (50:50) an increase in linear dimension could be as large as about 50% (see Table III.8), a hydrogel contact lens of such material having a final central thickness of, say 0.15 mm will require its xerogel (i.e. dehydrated lens) to be 0.1 mm thick. This means that for a hydrogel contact lens with a larger linear swell factor, it will be more difficult to make a sufficiently thin dehydrated lens if the final hydrated lens is to have a central thickness of, say 0.1 mm.

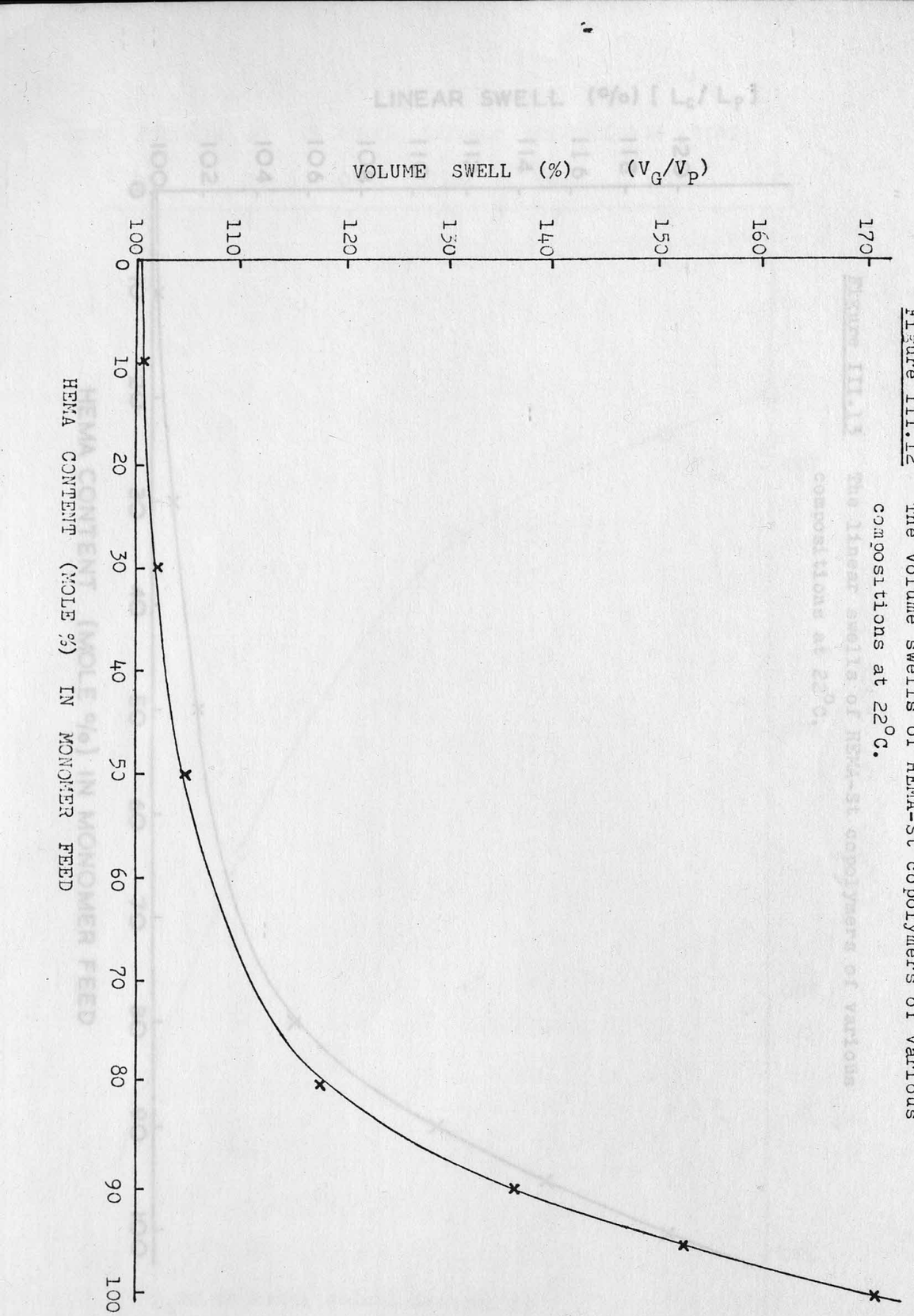


Figure III.12 The volume swells of HEMA-St copolymers of various compositions at 22°C.

Figure III.13 The linear swells of HEMA-St copolymers of various compositions at 22°C.

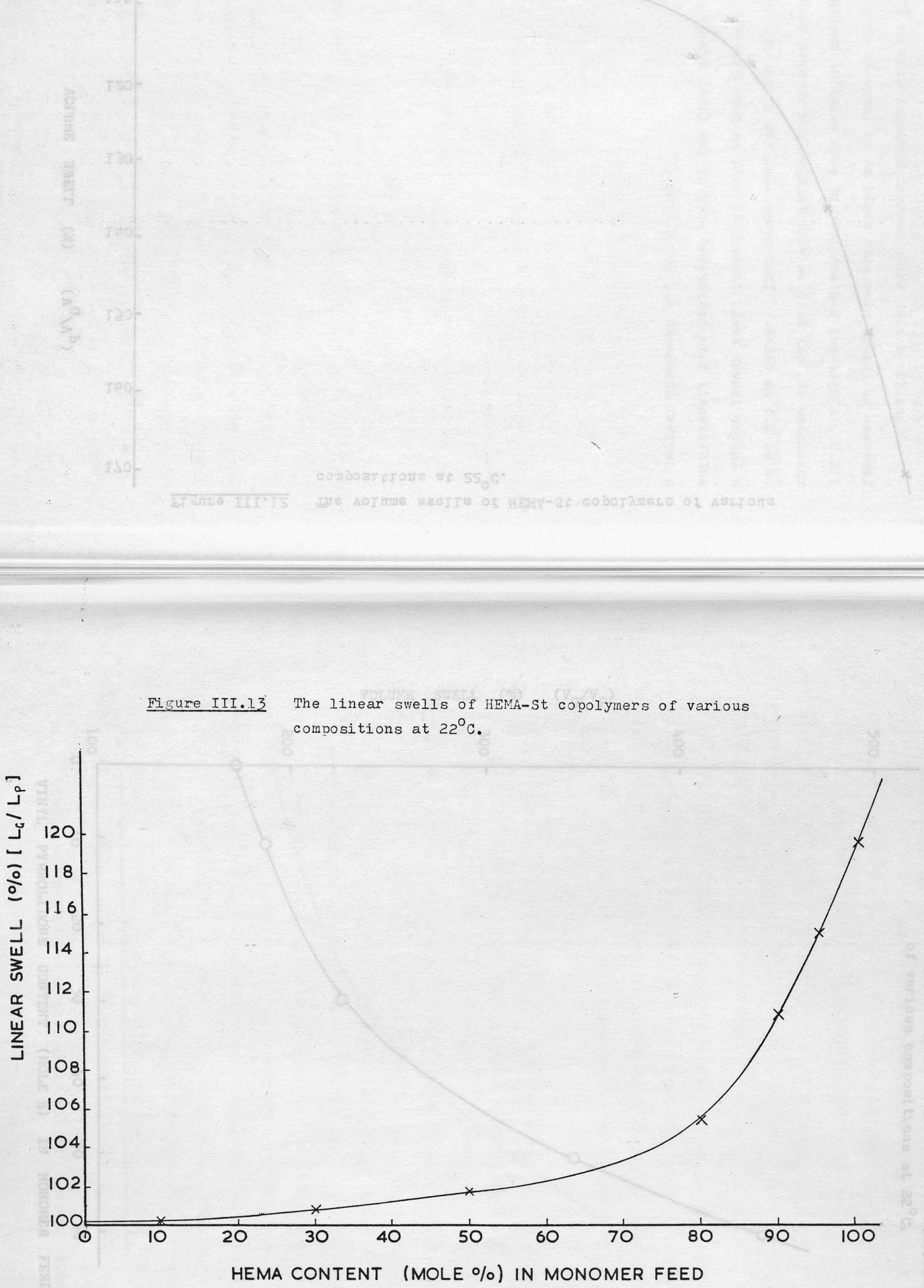
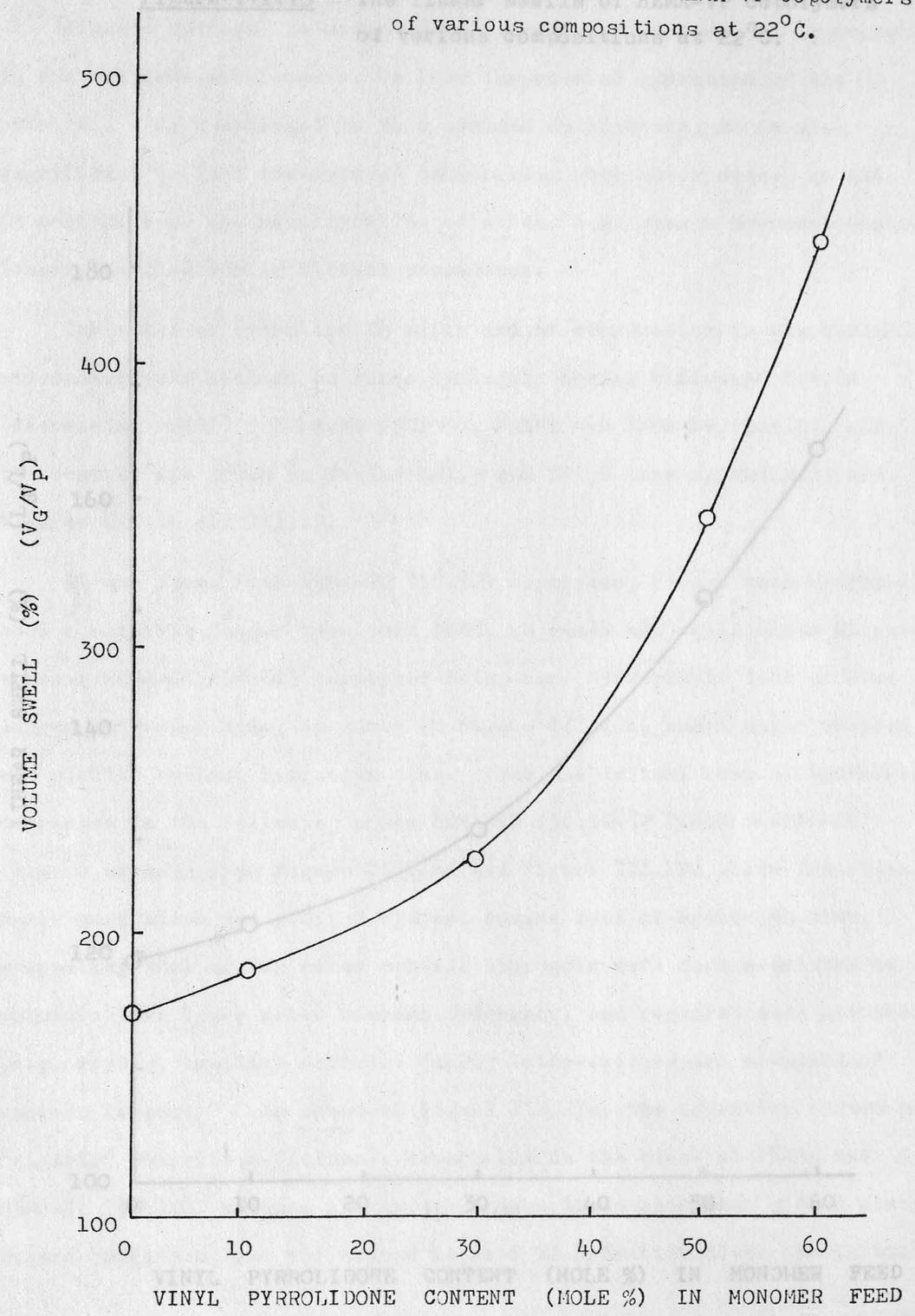


Figure III.14 The volume swells of HEMA-VP copolymers of various compositions at 22°C.



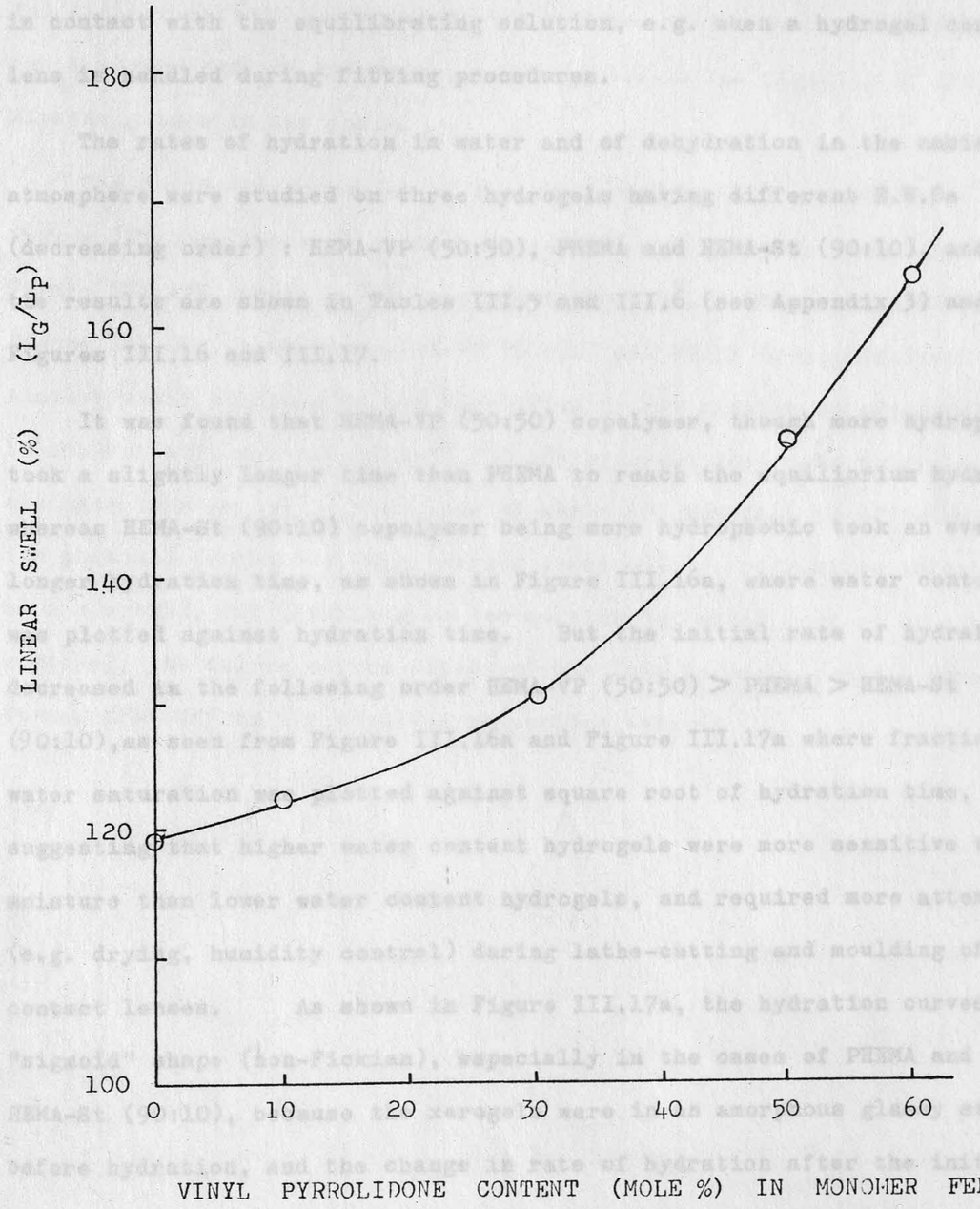


Figure III.15 The linear swells of HEMA-VP copolymers of various compositions at 22°C.

III.4. RESULTS AND DISCUSSIONS ON RATES OF HYDRATION AND DEHYDRATION OF SOME HYDROGELS

Since a hydrogel is obtained by hydrating a xerogel, it is important, in contact lens manufacture, to know the rate of hydration of the hydrogel. As a hydrogel is in a dynamic equilibrium, it is also significant to know the rate of dehydration when the hydrogel is not in contact with the equilibrating solution, e.g. when a hydrogel contact lens is handled during fitting procedures.

The rates of hydration in water and of dehydration in the ambient atmosphere were studied on three hydrogels having different E.W.Cs (decreasing order) : HEMA-VP (50:50), PHEMA and HEMA-St (90:10), and the results are shown in Tables III.5 and III.6 (see Appendix 3) and Figures III.16 and III.17.

It was found that HEMA-VP (50:50) copolymer, though more hydrophilic, took a slightly longer time than PHEMA to reach the equilibrium hydration whereas HEMA-St (90:10) copolymer being more hydrophobic took an even longer hydration time, as shown in Figure III.16a, where water content was plotted against hydration time. But the initial rate of hydration decreased in the following order HEMA-VP (50:50) > PHEMA > HEMA-St (90:10),as seen from Figure III.16a and Figure III.17a where fractional water saturation was plotted against square root of hydration time, suggesting that higher water content hydrogels were more sensitive to moisture than lower water content hydrogels, and required more attention (e.g. drying, humidity control) during lathe-cutting and moulding of contact lenses. As shown in Figure III.17a, the hydration curves are "sigmoid" shape (non-Fickian), especially in the cases of PHEMA and HEMA-St (90:10), because the xerogels were in an amorphous glassy state before hydration, and the change in rate of hydration after the initial

hydration seemed to indicate the transition from a glassy state to a rubbery state due to the plasticisation of the polymer chains by the absorbed water. The rates of hydration in the rubbery state were calculated for the three polymers from the plot $\ln\left[1 - \frac{(W_W)_t}{(W_W)_\infty}\right]$ vs. t, shown in Figure III.18, and the results are shown in Table III.9. (see Appendix 3). The results show that both PHEMA and HEMA-VP (50:50) had similar rate of hydration in the rubbery state, but HEMA-St (90:10) had a slower rate of hydration, possibly due to the rigidity of the styrene groups in the chain.

The dehydration studies of the three hydrogels are shown in Figure III.16b, where water content was plotted vs. time, and also in Figure III.17b, where $\frac{(W_W)_t}{(W_W)_o}$ was plotted vs. $t^{\frac{1}{2}}$. It can be seen from Figure III.17b, that both HEMA-VP (50:50) and PHEMA hydrogels lost similar water contents in the first five or ten minutes on exposure to ambient air, whereas HEMA-St (90:10) hydrogel lost less water in the same time period. This loss of water is important in affecting the physical properties (e.g. oxygen permeability, refractive index) of a hydrogel, but depending on the equilibrium water content of a hydrogel, the degree of the effect of the loss in water content (due to dehydration) on its physical properties varied.

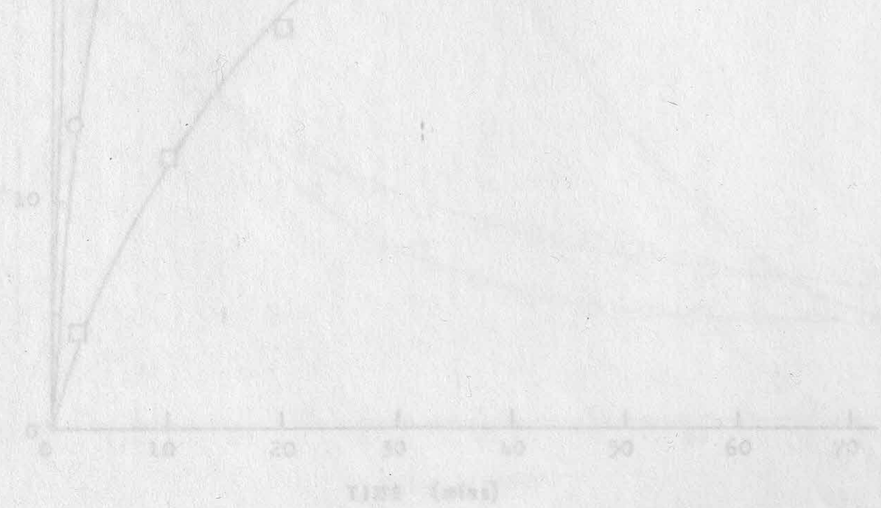


Figure III.16a The water contents of some materials at various hydration times during hydration in water at 22°C.

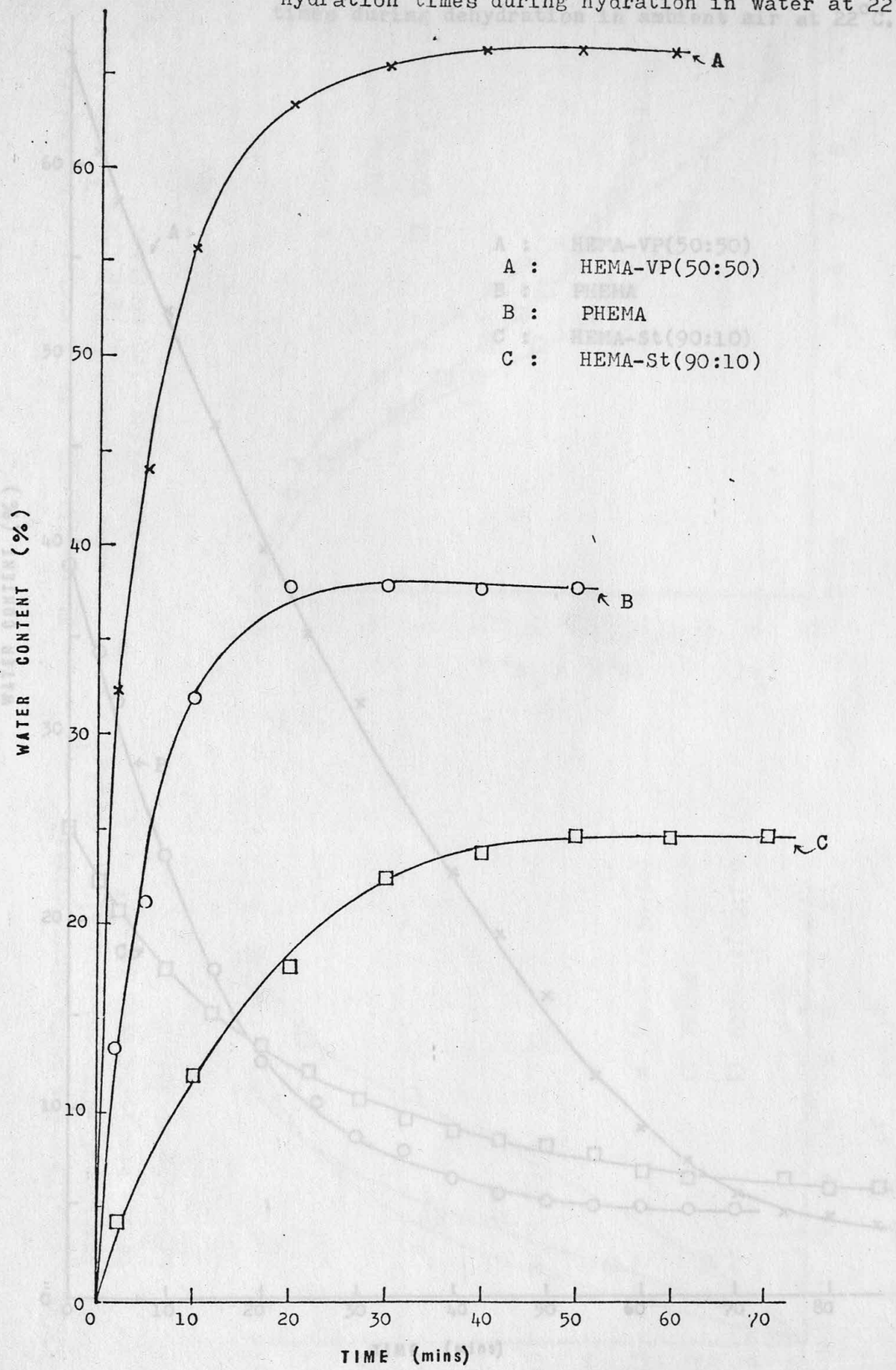


Figure III.16b The water contents of some materials at various times during dehydration in ambient air at 22°C.

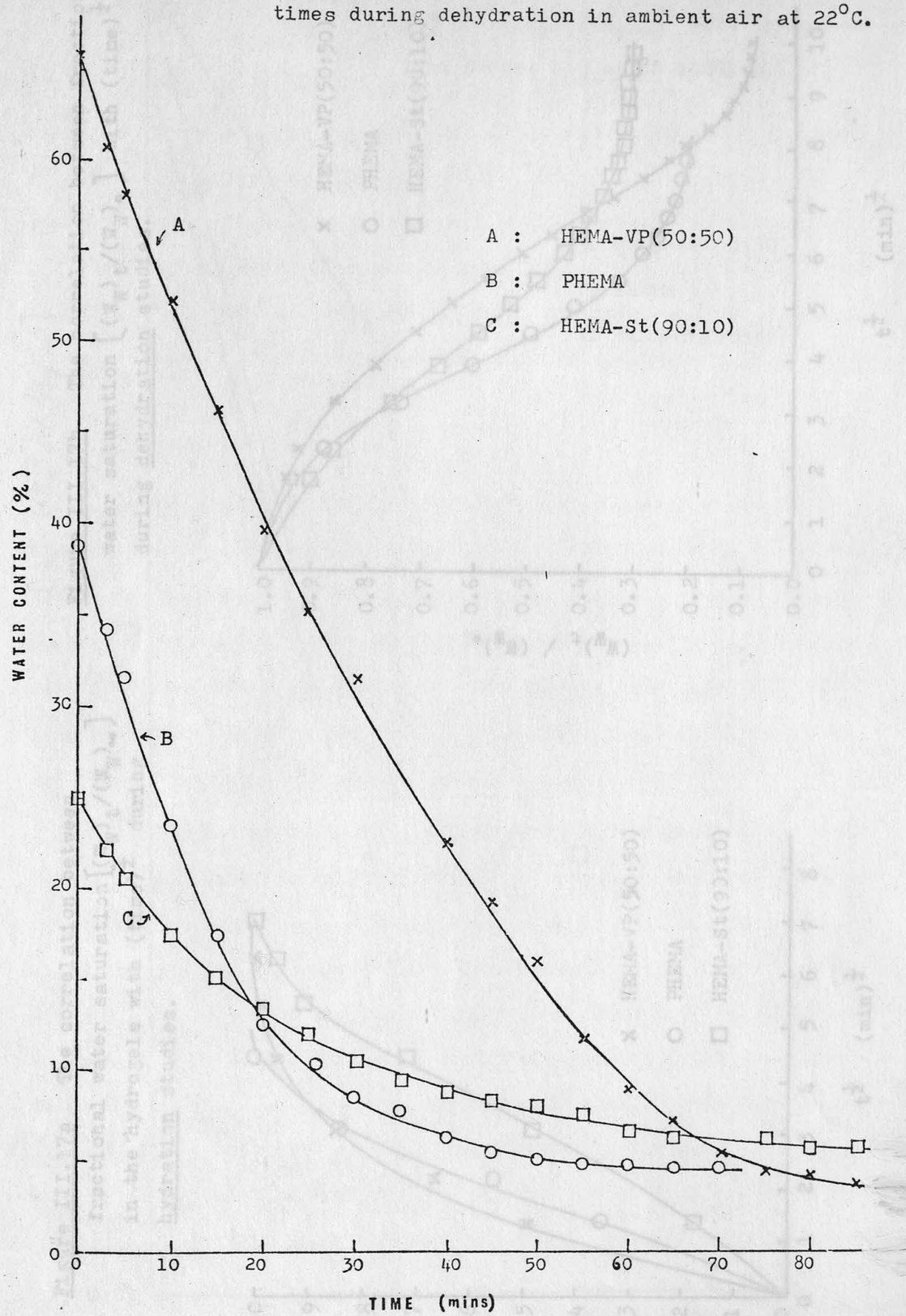


Figure III.17a The correlation between fractional water saturation $\left[(W_W)_t/(W_W)_\infty\right]$ in the hydrogels with $(\text{time})^{\frac{1}{2}}$ during hydration studies.

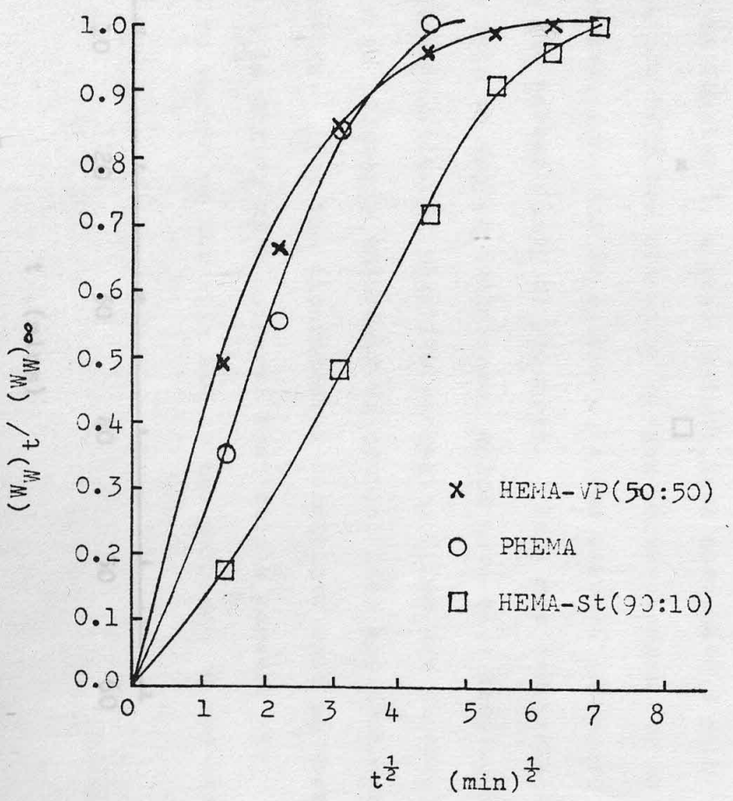
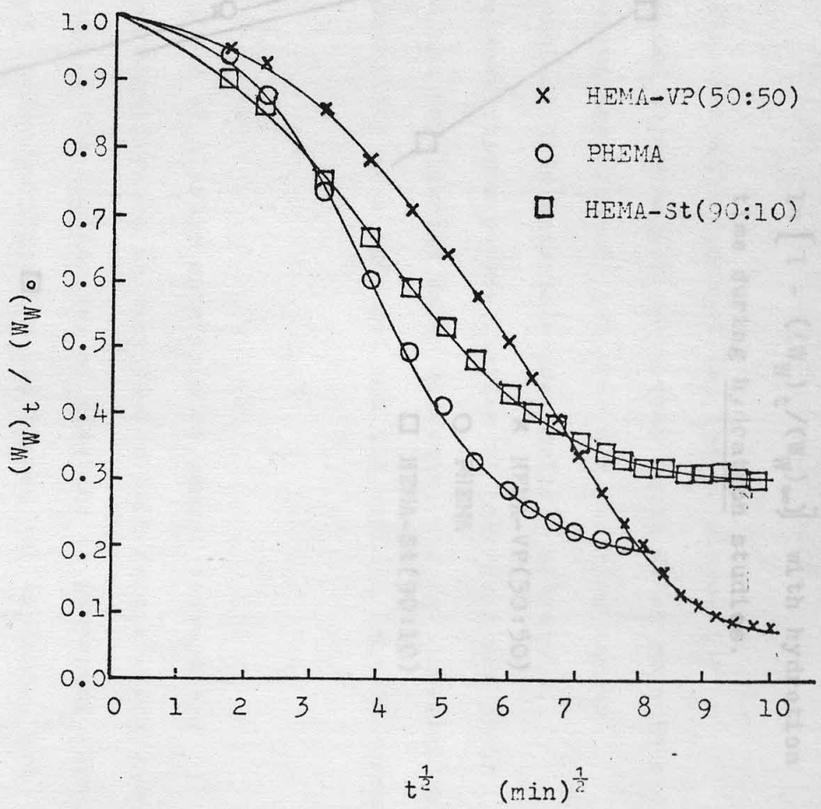


Figure III.17b The correlation between fractional water saturation $\left[(W_W)_t/(W_W)_o\right]$ with $(\text{time})^{\frac{1}{2}}$ during dehydration studies.



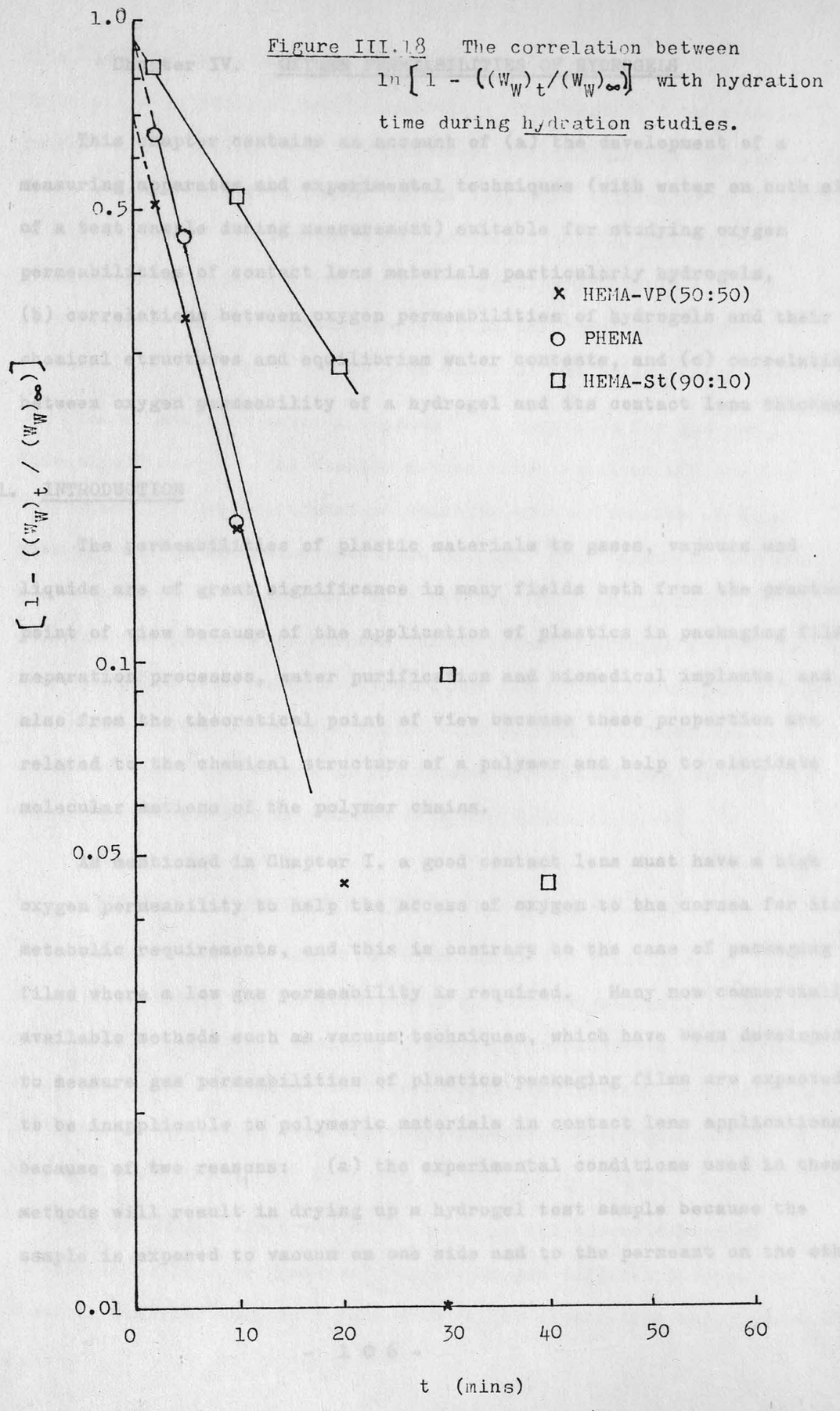
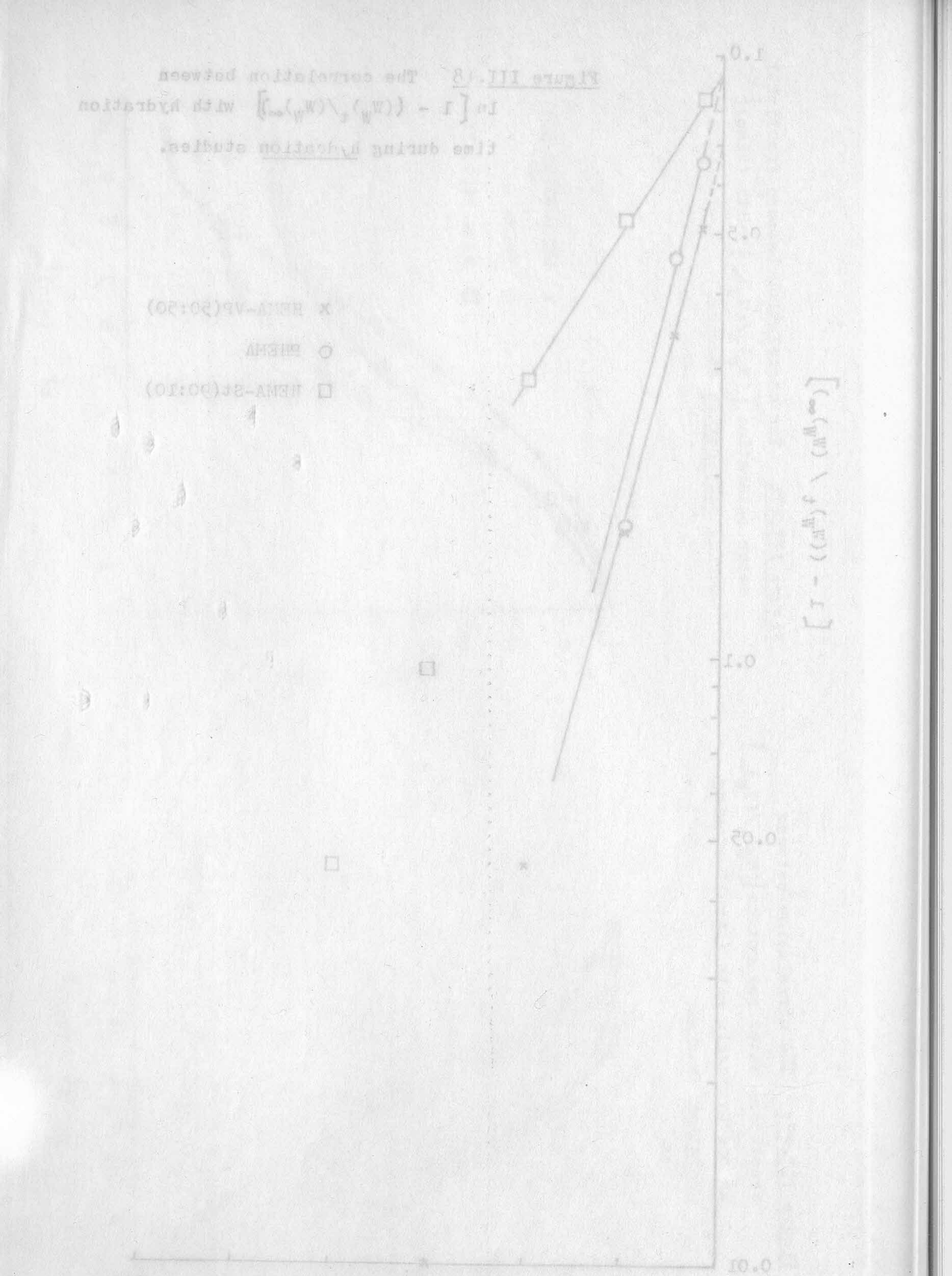


Figure III.18 The correlation between $\ln\left[1 - ((W_W)_t/(W_W)_\infty)\right]$ with hydration time during hydration studies.



Chapter IV. OXYGEN PERMEABILITIES OF HYDROGELS

This chapter contains an account of (a) the development of a measuring apparatus and experimental techniques (with water on both sides of a test sample during measurement) suitable for studying oxygen permeabilities of contact lens materials particularly hydrogels, (b) correlations between oxygen permeabilities of hydrogels and their chemical structures and equilibrium water contents, and (c) correlation between oxygen permeability of a hydrogel and its contact lens thickness.

IV.1. INTRODUCTION

The permeabilities of plastic materials to gases, vapours and liquids are of great significance in many fields both from the practical point of view because of the application of plastics in packaging films, separation processes, water purification and biomedical implants, and also from the theoretical point of view because these properties are related to the chemical structure of a polymer and help to elucidate molecular motions of the polymer chains.

As mentioned in Chapter I, a good contact lens must have a high oxygen permeability to help the access of oxygen to the cornea for its metabolic requirements, and this is contrary to the case of packaging films where a low gas permeability is required. Many now commercially available methods such as vacuum techniques, which have been developed to measure gas permeabilities of plastics packaging films are expected to be inapplicable to polymeric materials in contact lens applications because of two reasons: (a) the experimental conditions used in these methods will result in drying up a hydrogel test sample because the sample is exposed to vacuum on one side and to the permeant on the other

side, and (b) these methods cannot measure oxygen "dissolved" in water. To obtain a reliable oxygen permeability of a contact lens material in contact lens applications, it is logical to simulate the conditions in which a contact lens is placed in the eye, i.e. with water on both sides of the polymer. This means that in contact lens applications it is more practical to consider the "dissolved" oxygen permeability (i.e. with water on both sides of the polymer sample) than the "gaseous" oxygen permeability (i.e. with gas on both sides of the polymer sample).

There have been several methods[211] suggested for the analysis of "dissolved" oxygen: (a) Winkler method which involves a titration technique, (b) Gasometric method which is another version of Winkler method, and (c) membraned electrode method. The first two techniques are considered not applicable here because they involve tedious experimental procedures, whereas the membraned electrode method, on the other hand, is more straight forward and can follow continuous change in oxygen concentration in a liquid or a solution. Therefore the membraned electrode method is the technique used in this work to study oxygen permeabilities of contact lens materials.

IV.1.A. Principle of Membraned Electrode (i.e. Oxygen Electrode)

A solid membraned electrode operates on the principles of polarographic analysis, and is essentially an electrolysis cell consisting of a metallic electrode (cathode), a reference electrode (anode) and an electrolyte, as well as a plastics membrane separating the electrolysis cell and the liquid or solution whose gaseous concentration is to be measured.

For an oxygen electrode which measures the concentration of "dissolved" oxygen or "gaseous" oxygen, oxygen diffuses through the plastics membrane to a very thin film of the electrolyte and then to the

cathode which is maintained at a negative voltage with respect to the reference electrode (i.e. anode). Electro-reduction of oxygen takes place at the cathode surface and a limiting current is thus produced which is linearly proportional to the oxygen concentration at the cathode surface. The chemical reactions occurring at the electrodes of a polarographic oxygen probe consisting of a metallic cathode (e.g. gold, platinum), a silver-silver chloride anode and a KCl electrolyte are

$$O_2 + 2H_2O + 4e^- \longrightarrow 4OH^- \qquad \text{(cathode)}$$

$$4Ag + 4Cl^- \longrightarrow 4AgCl + 4e^- \qquad \text{(anode)}$$

since the plastics membrane of the oxygen probe separates its electrolyte and the solution to be analysed, the membrane should be permeable to the oxygen being analysed and non-permeable to the other solutes of the electrolyte and the solution. Polythene and PTFE have been found suitable in this application. Being introduced by Clark(212) to measure continuously the oxygen tension in blood, polythene membraned electrodes have been modified and used to measure "dissolved" oxygen concentration in biological systems, in streams, in manufacturing processes and in other fields.

IV.1.B. Theory of Oxygen Permeability.

"Permeability" is defined as the ease of transmission of a penetrant (gas, vapour or liquid) through a resisting material without consideration of the actual mechanism involved. Since "permeability" has been used in many different disciplines and research fields, it is invalid to compare permeability data without considering the definition of permeability used.

The permeation of small molecules such as oxygen through a polymeric film can occur by different processes such as activated diffusion or capillary flow.

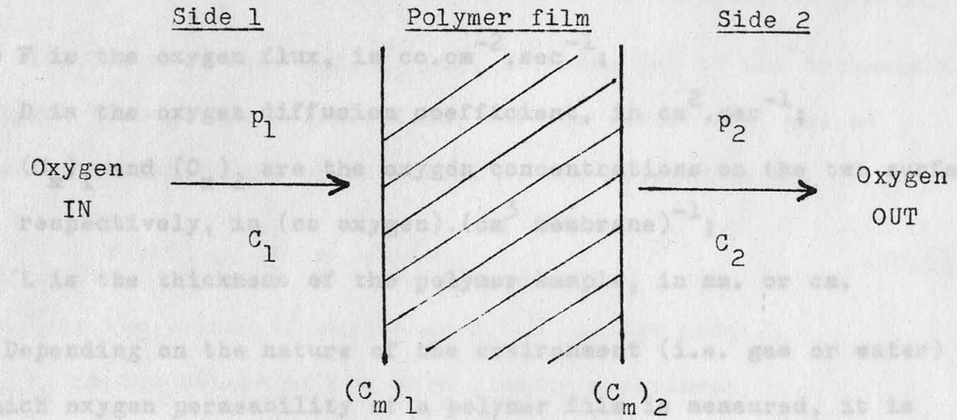
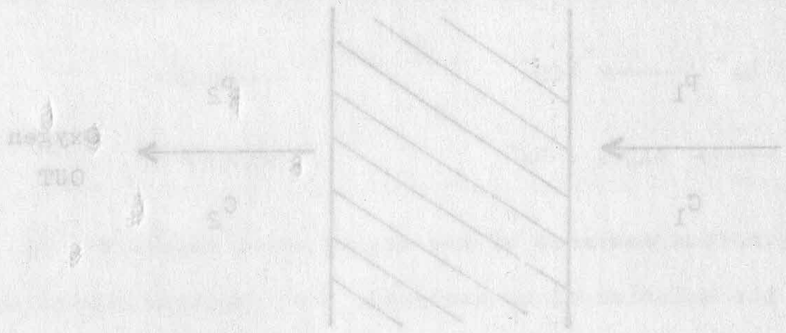


Figure IV.1 Oxygen permeation through polymer films.



For oxygen permeability by the molecular diffusion process, the oxygen flux (i.e. rate of oxygen transport per unit area) from side 1 to side 2 of the polymer film (see Figure IV.1) under steady-state conditions can be expressed by Fick's first law, as Equation IV.1

$$F = D \frac{\Delta C_m}{L} = D \frac{(C_m)_1 - (C_m)_2}{L} \qquad \text{(IV.1)}$$

where F is the oxygen flux, in $cc.cm^{-2}.sec^{-1}$;

D is the oxygen diffusion coefficient, in $cm^2.sec^{-1}$;

$(C_m)_1$ and $(C_m)_2$ are the oxygen concentrations on the two surfaces respectively, in (cc oxygen).$(cm^3$ membrane$)^{-1}$;

and L is the thickness of the polymer sample, in mm. or cm.

Depending on the nature of the environment (i.e. gas or water) in which oxygen permeability of a polymer film is measured, it is realistic to consider both "gaseous" and "dissolved" oxygen permeabilities.

IV.1.B.1. <u>"Gaseous" Oxygen Permeability</u> (symbolled "GO" Permeability)

If both sides of the polymer film are in the gas phase (symbolled G/G), then the oxygen concentration on each of the film surfaces can be related to the pressure of the gaseous oxygen in contact with that surface by Henry's law, i.e.

$$(C_m)_1 = Sp_1 \quad \text{and} \quad (C_m)_2 = Sp_2 \qquad \text{(IV.2)}$$

where S is the solubility coefficient of oxygen in the polymer,

p_1 and p_2 are the oxygen pressures in sides 1 and 2 of the polymer respectively,

Substitution of Equation IV.1 into Equation IV.2 gives Equations IV.3 and IV.4,

$$F = D \cdot S \frac{p_1 - p_2}{L} = P_g \frac{p_1 - p_2}{L} \qquad \text{(IV.3)}$$

where $$P_g = D \cdot S \qquad \text{(IV.4)}$$

and is the "GO" permeability coefficient of the polymer in question.

If p_1 is kept constant, p_2 will increase due to the transmission of oxygen through the polymer film, and the value of p_2 can be converted to the volume of oxygen at S.T.P. using Equation IV.5

$$V = V_2 \times \frac{p_2}{76} \times \frac{273}{(273+T)} \qquad \text{(IV.5)}$$

where V is the volume of oxygen at S.T.P. in the side 2,

V_2 is the volume of the side 2 under experiment,

T is the temperature of the test, in ^{o}C ,

and p_2 is the pressure of oxygen in the side 2, in cm. Hg.

$$\text{i.e.} \quad \frac{dV}{dt} = V_2 \times \frac{dp_2}{dt} \times \frac{1}{76} \times \frac{273}{(273+T)} \qquad \text{(IV.6)}$$

Equation IV.3 can be written as Equation IV.7 and substitution of Equation IV.6 into Equation IV.7 gives Equation IV.8

$$F = \frac{dV}{A.dt} = P_g \frac{p_1 - p_2}{L} \qquad \text{(IV.7)}$$

$$\frac{V_2}{A} \times \frac{dp_2}{dt} \times \frac{1}{76} \times \frac{273}{(273+T)} = \frac{P_g}{L} \times (p_1 - p_2) \qquad \text{(IV.8)}$$

Integration of Equation IV.8 gives Equation IV.9

$$\ln (p_1 - p_2) - \ln p_1 = - \frac{P_g}{L} \times \frac{A}{V_2} \times 76 \times \frac{(273+T)}{273} \times t \qquad \text{(IV.9)}$$

If k is the slope (negative) of the plot of $\ln (p_1 - p_2)$ vs. t, then

$$k = \frac{P_g}{L} \times \frac{A}{V_2} \times 76 \times \frac{(273+T)}{273}$$

so $$P_g = \frac{V_2}{A} \times \frac{1}{76} \times \frac{273}{(273+T)} \times kL \qquad \text{(IV.10)}$$

where P_g is the "GO" permeability coefficient, in [cc(S.T.P.).mm.cm^{-2}. sec^{-1}. cm Hg^{-1}]

V_2 in cc., A in cm^2, T in ^oC, k in sec^{-1}, and L in mm.

IV.1.B.2. <u>"Dissolved" Oxygen Permeability</u> (Symbolled "DO" Permeability)

If both sides of the polymer sample are water (symbolled W/W), as in the case of contact lens applications, then the concentrations of "dissolved" oxygen have to be considered in the derivation of "DO" permeability coefficient. It can be seen from Figure IV.1 that

$$C_1 = S_W \cdot p_1 \quad \text{and} \quad C_2 = S_W \cdot p_2 \qquad \text{(IV.11)}$$

where C_1 and C_2 are the "DO" concentrations in water in the sides 1 and 2 respectively, and

S_W is the solubility coefficient of oxygen in water.

The oxygen flux (F) can also be expressed by Equation IV.12

$$F = P \cdot \frac{C_1 - C_2}{L} \qquad \text{(IV.12)}$$

where P is the permeability coefficient in cm^2.sec^{-1}

F, C_1, C_2 and L are defined as earlier.

Substitution of Equation IV.11 into Equation IV.12 gives Equation IV.13 and Equation IV.14

$$F = P.S_W \cdot \frac{p_1 - p_2}{L}$$

$$= P_d \frac{p_1 - p_2}{L} \qquad \text{(IV.13)}$$

where $P_d = P.S_W$ (IV.14)

and P_d is called the "DO" permeability coefficient of the polymer in question.

Since the oxygen flux F can be alternatively written as Equation IV.15,

$$F = \frac{1}{A} \cdot \frac{dV}{dt} = \frac{1}{A} \left[V_2 \cdot \left(S_W \cdot \frac{dp_2}{dt} \right) \right] \qquad (IV.15)$$

substitution of Equation IV.13 into Equation IV.15 gives

$$\frac{dp_2}{dt} = \frac{P_d.A}{V_2 S_W L} (p_1 - p_2)$$

and integration of this gives Equation IV.16

$$\ln (p_1 - p_2) - \ln p_1 = -\frac{P_d.A}{V_2 S_W L} \cdot t \qquad (IV.16)$$

A plot of $\ln (p_1 - p_2)$ vs. t gives a slope k (negative), where

$$k = \frac{P_d.A}{V_2 S_W L} \quad ,$$

and so

$$P_d = \frac{V_2 \cdot S_W}{A} (kL) \qquad (IV.17)$$

where P_d is in [cc(S.T.P).mm.cm^{-2}.sec^{-1}.cm Hg^{-1}],

V_2 in cc., A in cm^2, S_W in [cc(S.T.P) O_2. (cm^3 H_2O)$^{-1}$. cm Hg^{-1}],

k in sec^{-1} and L in mm.

By comparing Equation IV.1 and Equation IV.12, we o btain

$$P \times \frac{C_1 - C_2}{L} = D \times \frac{(C_m)_1 - (C_m)_2}{L} \qquad (IV.18)$$

If a distribution coefficient (or partition ratio), σ, of oxygen between the polymer sample and water is defined as

$$\sigma = \frac{(C_m)_1}{C_1} = \frac{(C_m)_2}{C_2}$$

that is, $(C_m)_1 = \sigma C_1$ and $(C_m)_2 = \sigma C_2$ (IV.19)

then by substituting Equation IV.19 into Equation IV.18 we obtain

$$P = D.\sigma \qquad \text{(IV.20)}$$

or $$P_d = D.\sigma.S_W \qquad \text{(IV.21)}$$

IV.2. EXPERIMENTAL

This section consists of (a) development of a measuring apparatus and establishment of experimental techniques to enable the oxygen permeability of a polymer film particularly a hydrogel material to be measured in the "dissolved" oxygen conditions, and (b) measurements of "DO" permeability coefficients of most hydrogel materials made in this work with an attempt to correlate with their structures.

IV.2.A. Oxygen Permeability Measurements with the Davenport Gas Permeability Apparatus

The Davenport Gas Permeability Apparatus is used commercially to measure gas permeability of a plastics film (or a rubber sheet) in gas/gas (G/G) conditions, This apparatus was used here to study oxygen permeabilities of some hydrophobic plastics materials (potential contact lens materials) under "GO" conditions, and also to check validity of oxygen permeability measurements of the same materials in those apparatus involving a membraned electrode (Section IV.2.B) under the same "GO" conditions.

IV.2.A.1. Description of the Apparatus

The Davenport Gas Permeability Apparatus (Figure IV.2 and Plate IV.1 in Appendix 6) consists of a stainless steel permeability cell connected through a capillary tube to a manometer and a vacuum system, and is fitted to a vertical mounting.

The permeation cell unit has two parts which contain a rubber "O" ring to form a seal when they are clamped together by six vertical studs with the nuts. The lower part of the cell is fitted with a changeable insert so that the volume of the free space in this part of the cell is

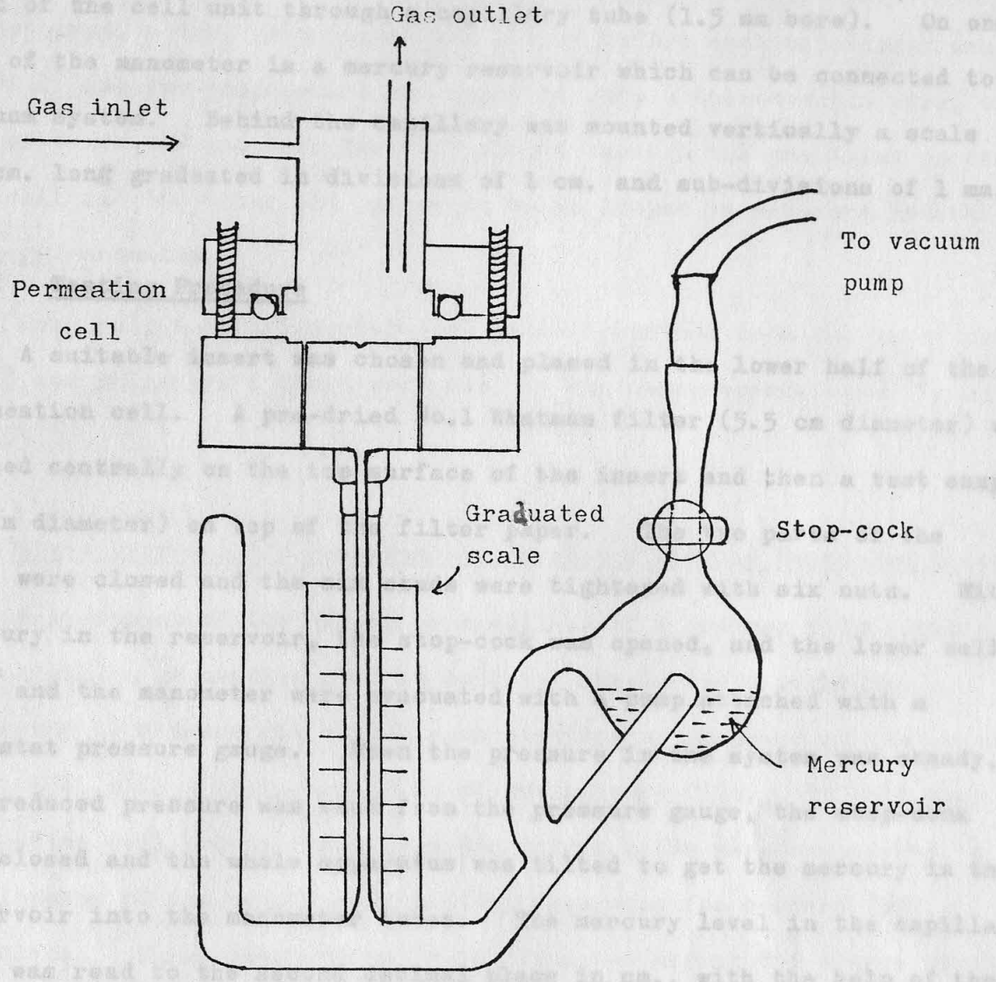
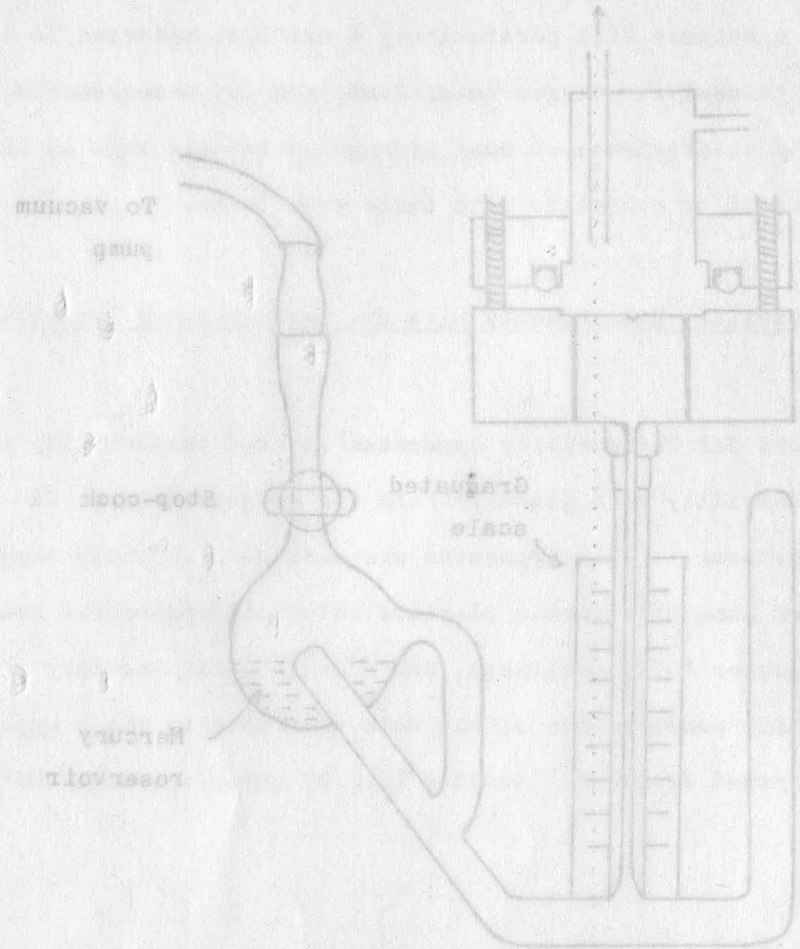


Figure IV.2 The Davenport Gas Permeability Apparatus.



20, 15, 10 or 5 mls.

The manometer unit consists of a manometer connected to the lower part of the cell unit through a capillary tube (1.5 mm bore). On one end of the manometer is a mercury reservoir which can be connected to a vacuum system. Behind the capillary was mounted vertically a scale 10 cm. long graduated in divisions of 1 cm. and sub-divisions of 1 mm.

IV.2.A.2. Testing Procedure

A suitable insert was chosen and placed in the lower half of the permeation cell. A pre-dried No.1 Whatman filter (5.5 cm diameter) was placed centrally on the top surface of the insert and then a test sample (7 cm diameter) on top of the filter paper. The two parts of the cell were closed and the six studs were tightened with six nuts. With mercury in the reservoir, the stop-cock was opened, and the lower cell unit and the manometer were evacuated with a pump attached with a Vacustat pressure gauge. When the pressure in the system was steady, the reduced pressure was read from the pressure gauge, the stop-cock was closed and the whole apparatus was tilted to get the mercury in the reservoir into the manometer tubes. The mercury level in the capillary tube was read to the second decimal place in cm., with the help of the graduated scale behind it. Oxygen was passed from a cylinder into the top chamber of the cell and a stop-watch was immediately started. The mercury level in the capillary tube fell due to oxygen transmission through the test sample, and the readings were taken at time intervals. The height of the mercury was plotted against time, and the measurement was carried on until six consecutive readings showed a linear correlation. The gas was then switched off and the mercury was brought back to the reservoir before the stop-cock was opened again to evacuate the system for the next run.

In order to do oxygen permeability measurements at temperatures other than the room temperature, a heating tape was coiled on the cell unit and the heating was controlled with a pre-calibrated thermostat unit. If required, a test temperature was preset before each experiment was started, and the temperature was measured with a thermocouple wire, one end of which was put near the test sample through the gas inlet on the top cell and the other end connected to an Isopad temperature indicator or a galvanometer.

Only poly(4-methyl pent-1-ene) sheet (obtained from the Metal Box Ltd.) and polystyrene sheet were used in the measurements at 20°C, 25°C and 34°C.

IV.2.B. Oxygen Permeability Measurements Involving a Solid Membraned Electrode

Measurement of the oxygen permeability of a polymer film requires essentially a permeation cell to hold the sample and a membraned electrode to measure the oxygen concentration or pressure on the out-flux side (i.e. side 2 of Figure IV.1) of the polymer film. So both the permeation cell and the polarographic oxygen probe were considered in the development of the oxygen permeability apparatus for contact lens materials. Three combinations were attempted during the course of this work, and they are described as follows.

IV.2.B.1. Use of a Modified E.I.L. Oxygen Sensor

This piece of oxygen permeability apparatus was inherited from the Department of Ophthalmic Optics. It consists of an E.I.L. oxygen sensor (electrode system) modified with an attachment of a small permeation cell to hold a polymer sample in question. The oxygen sensor was supplemented with a voltage polarising unit and an electrometer.

IV.2.B.1a. Description of the Oxygen Sensor, Voltage Polarising Unit and Electrometer

The oxygen sensor, voltage polarising unit and electrometer which were required in the measurements were bought from the Electronic Instruments Ltd., (E.I.L.).

As shown in Figure IV.3 and Plate IV.2 (see Appendix 6), the E.I.L. oxygen sensor consists of a platinum cathode (C) (E.I.L. Part No.19193) and a calomel anode (A) (Part No.19159) immersed in an electrolyte in a perspex electrode jacket, with a thin polythene membrane stretched over the tips of both the platinum electrode and the jacket, and fitted with a rubber "O" ring.

The voltage polarising unit (model D48A), which is battery operated (1.5 V) provides a variable polarising voltage of 0.5 - 0.9 V to the oxygen sensor and converts the current output due to reduction of oxygen at the cathode into a voltage signal suitable for use with a Vibron Electrometer. The polarising unit is equipped with two ranges 0 - 150 mm Hg and 0 - 760 mm Hg, and variable sensitivity to allow for varying current outputs from the electrodes.

The Vibron electrometer (model 33B-2) can measure various ranges of d.c. potentials from 0 - 10 mV to 0 - 1000 mV of either polarity. This instrument has a zero stability within 0.1 mV over a 12-hour period.

IV.2.B.1b. Description of the Permeation Cell

The small permeation cell made to attach to the E.I.L. oxygen sensor consists of three parts (made of brass) as shown in Figure IV.3. The part (E) was fixed to the jacket (J) of the oxygen sensor, and the part (F) could be fitted to the part (E) so that it sits on a rubber "O" ring against the tip of the perspex jacket, and the "hole" (H) of the

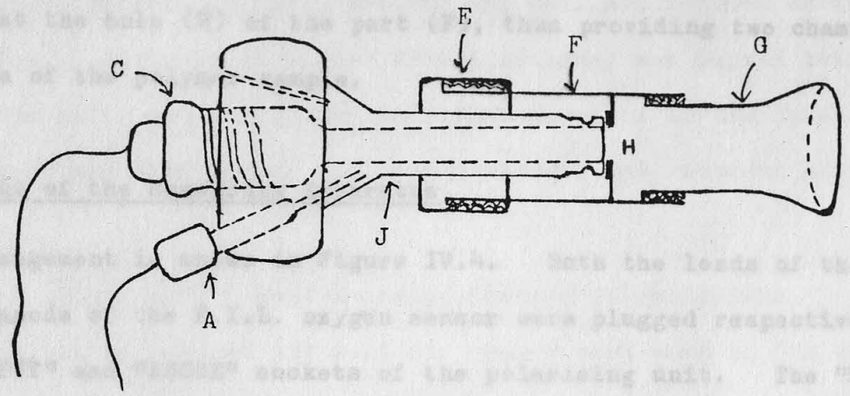


Figure IV.3 The E.I.L. polarographic oxygen sensor modified with a small oxygen permeation cell.

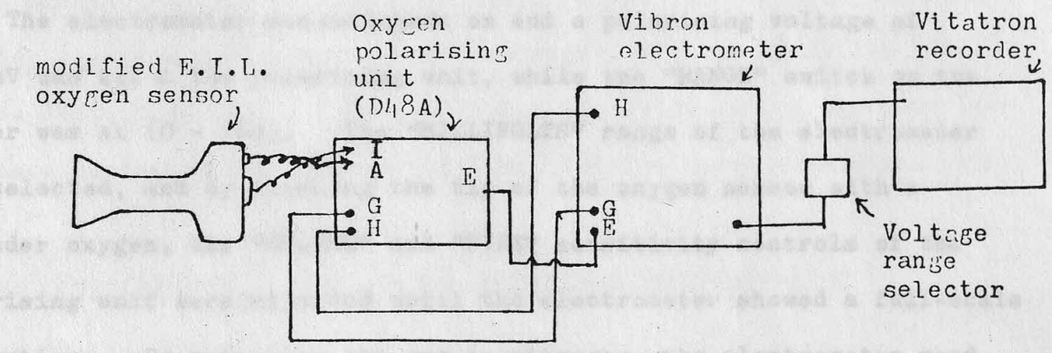
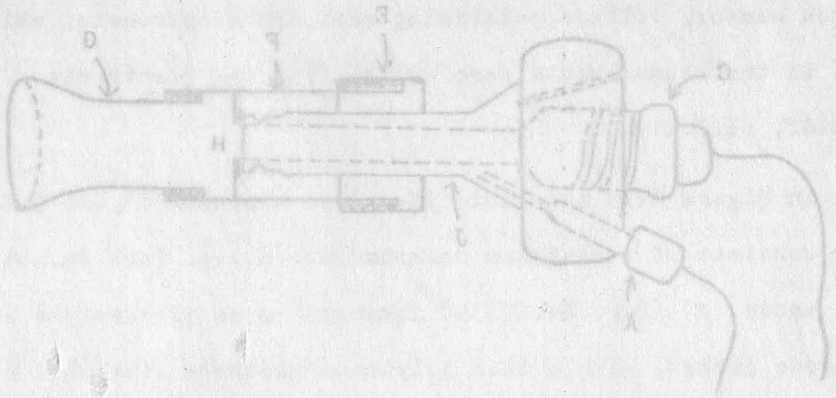
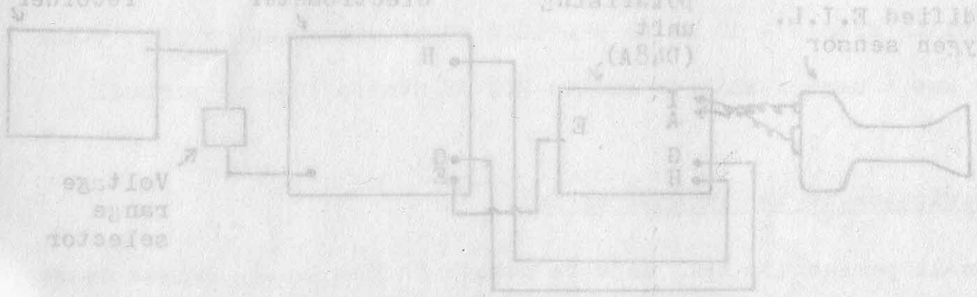


Figure IV.4 Layout of the oxygen permeability measuring apparatus involving the modified E.I.L. electrode system.





part (F) is just levelled with the tip of the cathode. The third part (G) could be screwed on to the part (F) to hold any polymer film placed against the hole (H) of the part (F), thus providing two chambers on both sides of the polymar sample.

IV.2.B.1c. Layout of the Measuring Apparatus

The arrangement is shown in Figure IV.4. Both the leads of the cathode and anode of the E.I.L. oxygen sensor were plugged respectively into the "INPUT" and "ANODE" sockets of the polarising unit. The "HIGH" and "GUARD" leads of the latter were connected respectively to the "HIGH" and "GUARD" sockets of the electrometer. Also the "EARTH" sockets of both the polarising unit and electrometer were joined together with a two-plug lead. The "INPUT POLARITY" switch on the electrometer was set at "POSITIVE" and the "INPUT SCREEN" switched to the "LOW" position. The output of the electrometer was connected to a Vitatron recorder through a voltage range selector.

IV.2.B.1d. Calibration and Testing Procedure

The electrometer was switched on and a polarising voltage of 650 mV was set at the polarising unit, while the "RANGE" switch on the latter was at (0 - 760). The "MILLIVOLTS" range of the electrometer was selected, and by flushing the tip of the oxygen sensor with a cylinder oxygen, the "COARSE" and "FINE" sensitivity controls of the polarising unit were adjusted until the electrometer showed a full-scale deflection. By switching the gas to nitrogen, the electrometer read zero. The whole system was then allowed to stabilise for some time preferably overnight. Before any actual oxygen permeability measurement was made, the full-scale deflection was checked again.

The oxygen permeability measurements were carried out under gas /gas conditions only. A hydrophobic test sample was mounted on the lower half of the brass permeation cell, and the top half was screwed on to the test sample. A stream of oxygen from a cylinder was passed into the top half of the cell, and the change in oxygen pressure in the lower half of the cell was detected by the oxygen sensor, and recorded on the recorder.

Sheets of poly(4-methyl pent-1-ene), oriented polypropylene, ethylene propylene terpolymer and silicone rubber were used in the oxygen permeability measurements using this apparatus.

IV.2.B.2. Use of an I.M.I. Oxygen Sensor and a Large Permeation Cell.

Due to some disadvantages of the E.I.L. electrode system and the associated measuring apparatus, which will be discussed later, another oxygen permeability measuring apparatus was considered and tried. This piece of apparatus consists of an I.M.I. polarographic oxygen sensor (I.M.I. No.3601A) with an associated I.M.I. Continuous Oxygen Monitor (both purchased from the Becton Dickinson U.K. Ltd.) and of a "home-made" large permeation cell. Since the I.M.I. oxygen sensor and oxygen monitor together with experimental techniques will be described later in the following section (Section IV.2.B.3), no intention was made here to describe this method except the design of the large permeation cell.

IV.2.B.2a. Descriptions of the Oxygen Sensor and Oxygen Monitor

See Section IV.2.B.3a.

IV.2.B.2b. Description of the Large Permeation Cell

The large permeation cell used here was made of perspex and its design is shown in Figure IV.5.

The principles of the design of the large permeation cell is

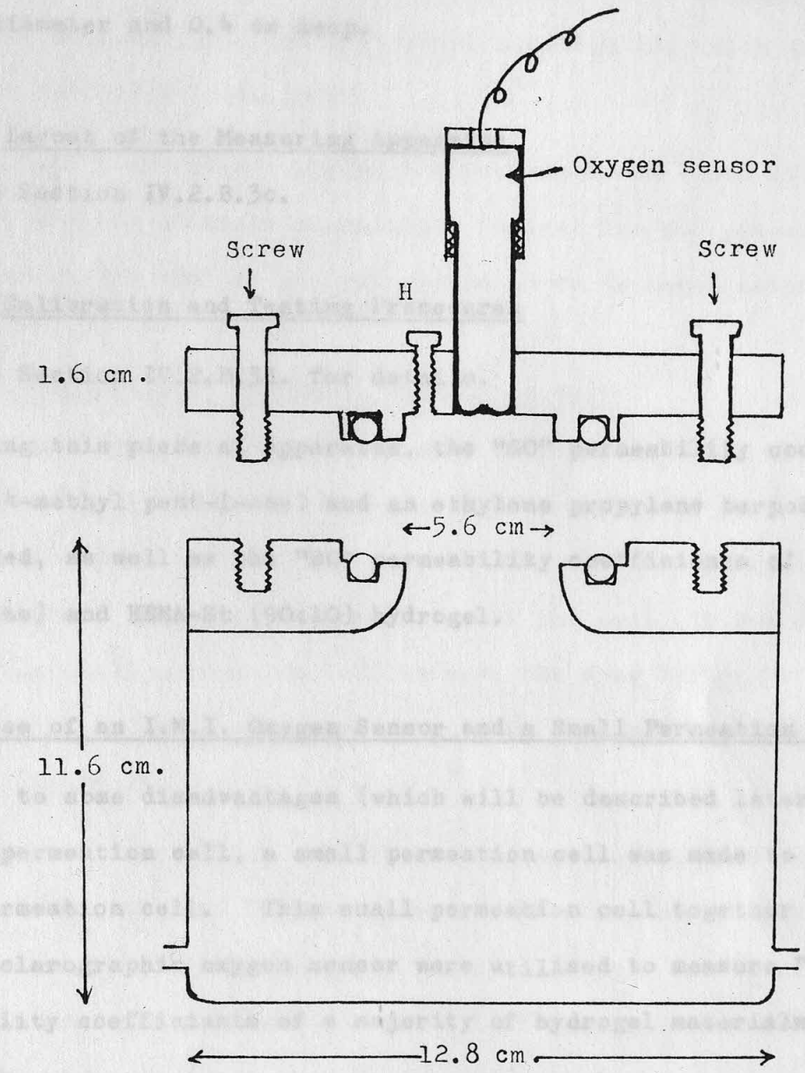
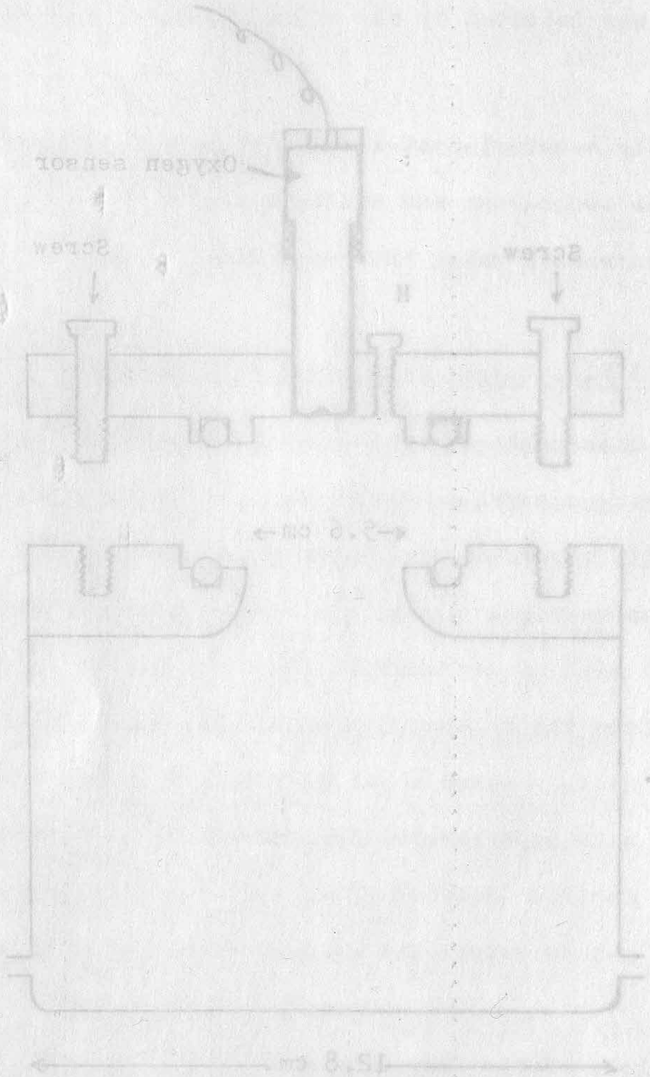


Figure IV.5 The large oxygen permeability cell.



similar to that of the small permeation cell, which will be mentioned in Section IV.2.B.3b., except that the large permeation cell is 12.8 cm. in diameter, 13.2 cm. tall, and takes 7.6 cm. diameter test sample disc between the two rubber "O" rings (internal diameter 6.2 cm and external diameter 7.4 cm.), giving the top half of the cell a volume of about 5.6 cm diameter and 0.4 cm deep.

IV.2.B.2c. Layout of the Measuring Apparatus

See Section IV.2.B.3c.

IV.2.B.2d. Calibration and Testing Procedures

See Section IV.2.B.3d. for details.

Using this piece of apparatus, the "GO" permeability coefficients of poly(4-methyl pent-1-ene) and an ethylene propylene terpolymer were determined, as well as the "DO" permeability coefficients of poly(4-methyl pent-1-ene) and HEMA-St (90:10) hydrogel.

IV.2.B.3. Use of an I.M.I. Oxygen Sensor and a Small Permeation Cell.

Due to some disadvantages (which will be described later) of using a large permeation cell, a small permeation cell was made to replace the large permeation cell. This small permeation cell together with the I.M.I. polarographic oxygen sensor were utilised to measure "DO" permeability coefficients of a majority of hydrogel materials made in this work.

IV.2.B.3a. Descriptions of the I.M.I. Oxygen Sensor and Oxygen Monitor

The I.M.I. oxygen sensor consists of a gold cathode insulated from a tubular silver anode by an epoxy casting, and of a disposable Teflon

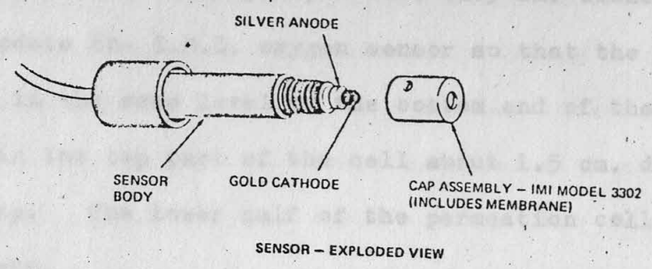
charge cap containing a Teflon membrane and a KCl electrolyte (see Figure IV.6 and Plate IV.4 in Appendix 6). The arrangements are such that both the cathode and the anode are fitted firmly against the Teflon membrane and electrochemically connected by a layer of the electrolyte. This I.M.I. oxygen sensor is smaller than the E.I.L. oxygen sensor used in Section IV.2.B.1, and has a disposable charge cap which is packaged in a "Charge Kit" (I.M.I. No.3302).

The I.M.I. continuous oxygen monitor (shown in Plate IV.4) is A.C. operated, and has adequate sensitivity control for the present work; the scales on the monitor are calibrated in mm Hg and % saturation.

IV.2.B.3b. Description of the Small Permeation Cell

The small permeation cell was made of perspex, which is transparent and impermeable to oxygen. The latter property is of particular importance because the cell was used to study oxygen permeabilities of polymer films. Owing to the smallness of the cell, it was difficult to machine the small permeation cell to have the same design as the large cell, and so the small cell was designed in a slightly different way, as shown in Figure IV.7.

This small permeation cell,which was constructed in a cylindrical shape (8.8 cm. tall and 5.1 cm. diameter) consists of two parts each having a rubber "O" ring (internal diameter 1.8 cm. and external diameter 2.1 cm) embedded in a groove so that a test sample disc (about 2.0 cm. diameter) can be placed on the flanges (F_B) (2.5 mm. wide) and clamped tight between the two "O" rings by six screws holding the two parts of the cell together. The top part of the cell was equipped with a hole (H) (which could be sealed with a nylon screw) for introducing a nitrogen-



SENSOR – EXPLODED VIEW

Figure IV.6 The I.M.I. polarographic oxygen sensor.

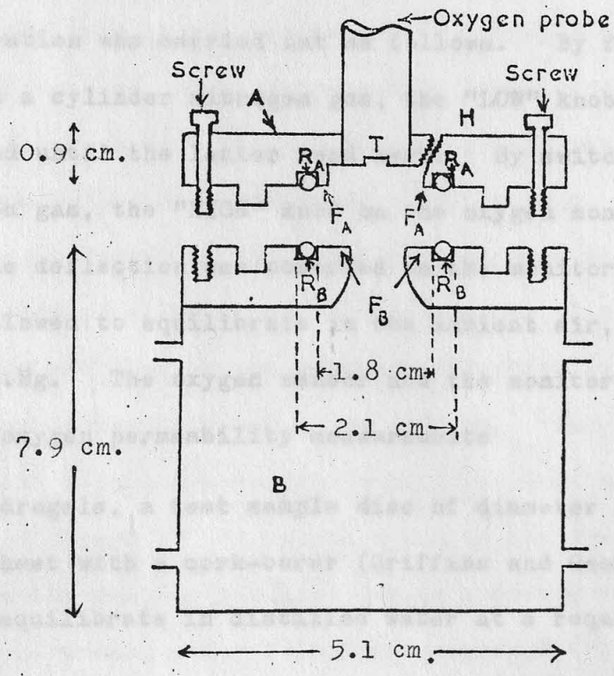
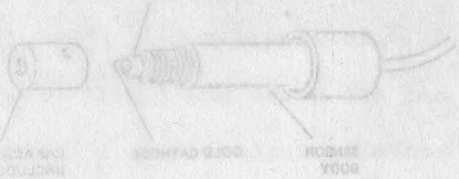
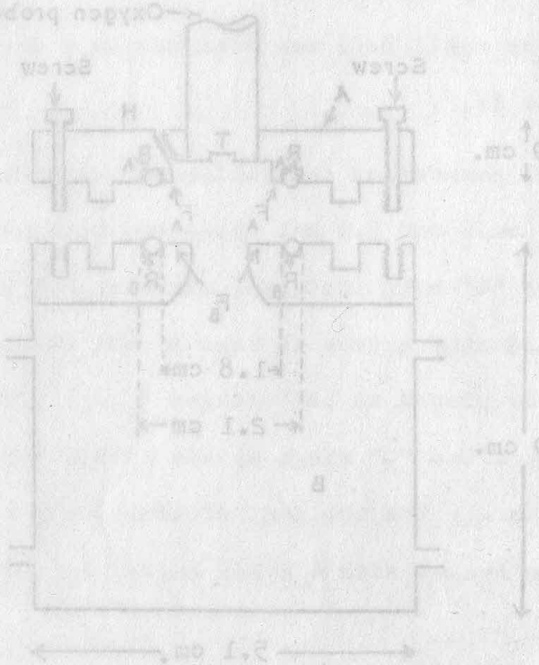


Figure IV.7 The small oxygen permeation cell.





saturated water, and also a vertical perspex tube (1.3 cm. diameter) with a screw cap to accommodate the I.M.I. oxygen sensor so that the tip of the oxygen sensor was in the same level as the bottom end of the perspex tube, giving a space in the top part of the cell about 1.5 cm. diameter and about 0.15 cm. deep. The lower half of the permeation cell had 4 water inlets and outlets.

IV.2.B.3c. Layout of the Measuring Apparatus

The layout of the oxygen permeability measuring apparatus is shown in Figure IV.8 and Plate IV.4 (see Appendix 6). The I.M.I. oxygen sensor was connected to the I.M.I. oxygen monitor, the output of which was then linked to a recorder (Model EPR-2TB, TOA Electronic Ltd.) The oxygen permeation cell was allowed to sit on a magnetic stirrer, and two water inlet and outlet of the cell were connected to a water bath thermostatted at a required temperature by an immersion thermostat unit.

IV.2.B.3d. Calibration and Testing Procedure

Calibration was carried out as follows. By flushing the oxygen sensor with a cylinder nitrogen gas, the "LOW" knob on the oxygen monitor was adjusted until the latter read zero. By switching the nitrogen gas to an oxygen gas, the "HIGH" knob on the oxygen monitor was adjusted until a full scale deflection was observed on the monitor. The oxygen sensor was then allowed to equilibrate in the ambient air, and the monitor should read 159 mm.Hg. The oxygen sensor and the monitor were then ready for use in the oxygen permeability measurements

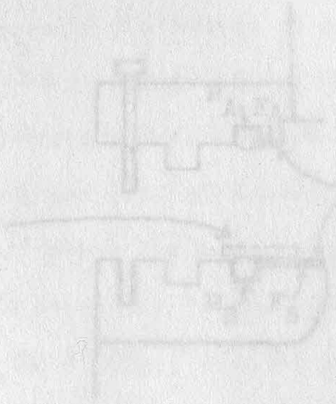
For hydrogels, a test sample disc of diameter 2.2. cm. was cut from a polymer sheet with a cork-borer (Griffins and George, size 15), and allowed to equilibrate in distilled water at a required test temperature

in a water bath for one week before it was cut again to a disc of diameter 1.8 cm or 1.95 cm. (depending on the sample mounting method which will be discussed in Section IV.3.A.4) with another cork-borer (sizes 12 and 13 respectively) and its thickness measured with a thickness gauge. For hydrophobic materials such as poly(4-methyl pent-1-ene) and polystyrene, a test sample disc of 1.8 cm or 1.95 cm. diameter was cut directly from a polymer sheet and its thickness measured with a micrometer.

The test sample was then clamped between the two halves of the permeation cell with a uniform pressure by gently adjusting the six screws to avoid twisting or rupturing the sample particularly in the case of soft hydrogels, while viewed from the bottom of the cell. The bottom half of the cell was then filled with a nitrogen-saturated water (which was made by bubbling a boiled distilled water overnight with a nitrogen gas) pre-equilibrated at the test temperature, keeping the sample in contact with the water, and the top half of the cell was flushed with a nitrogen gas which was introduced with a syringe needle through the hole (H). Any leak in the system due to improper clamping of test sample or presence of holes on the sample (e.g. accidental poking of the soft hydrogel sample with the tip of the syringe needle during the nitrogen flushing) could be detected at this stage. When the monitor read zero, a nitrogen-saturated water pre-equilibrated at the test temperature was then introduced slowly into the top chamber with a syringe through the hole while the entrance of the hole was blanketed with a nitrogen gas to prevent any atmospheric air going into it. Care was taken, while introducing the nitrogen-saturated water, to avoid any formation of bubbles at the trough of the oxygen sensor since bubbles were found later to give erroneous monitor readings during measurement. The volume of the nitrogen-saturated water introduced was read from the syringe scale

and the hole, still under the blanket of nitrogen gas, was then sealed with a nylon screw. The nitrogen-saturated water in the bottom half of the permeation cell was removed and then replaced with a continuously running air-saturated water thermostatted at the test temperature in the water-bath shown in Figure IV.8. The water in the bottom chamber was stirred with the magnetic stirrer, and the increase in oxygen concentration in the top chamber of the cell due to diffusion of oxygen from the bottom to the top chamber was indicated on the recorder which was set at a mV range of 100 and a chart speed of 60 mm/H.

When the measurement was completed, the test sample was unclamped and its thickness was again checked. Any water left in the trough of the oxygen sensor and the top half of the permeation cell was dried and oxygen sensor was allowed to equilibrate again in the ambient atmosphere for the next measurement. At least duplicate measurements were made for each material.



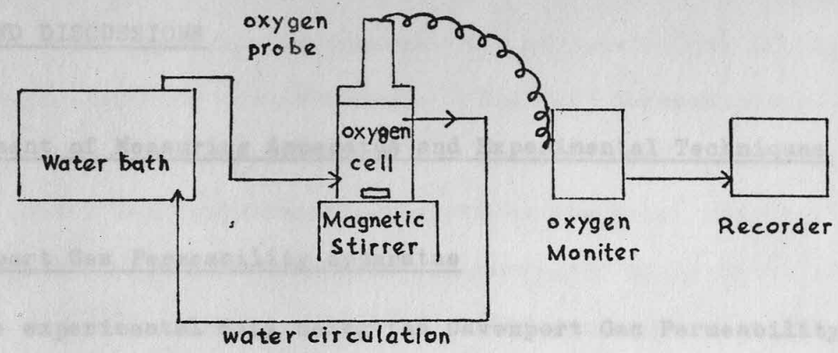


Figure IV.8 Layout of the oxygen permeability measuring apparatus involving the I.M.I. oxygen sensor.

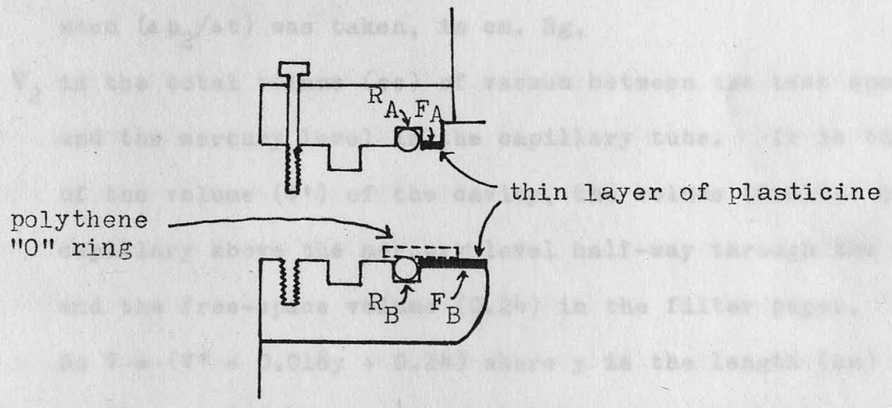


Figure IV.9 The applications of plasticine and a polythene "O" ring in the sample mounting (method 4).

IV.3. RESULTS AND DISCUSSIONS

IV.3.A. Development of Measuring Apparatus and Experimental Techniques

IV.3.A.1. Davenport Gas Permeability Apparatus

From the experimental data using the Davenport Gas Permeability Apparatus, the "GO" permeability coefficient of a material was calculated from Equation IV.8, i.e.

$$P_g = \left(\frac{273}{273+T} \times \frac{\Delta p_2}{\Delta t} \times \frac{1}{76} \times V_2 \right) \times \frac{1}{A} \times \frac{1}{(p_1 - p_2)} \times L \qquad \text{(IV.8)}$$

where T is the test temperature in ^{o}C,

$(\Delta p_2/\Delta t)$ is rate of pressure change in the capillary, in (cm Hg). sec^{-1},

A is the area of the specimen under test (23.77 cm^2 in this case),

$(p_1 - p_2)$ is the pressure difference across the sample at the time when $(\Delta p_2/\Delta t)$ was taken, in cm. Hg,

and V_2 is the total volume (cc) of vacuum between the test specimen and the mercury level in the capillary tube. It is the sum of the volume (V') of the cavity, the volume (0.018y) of the capillary above the mercury level half-way through the test and the free-space volume (0.24) in the filter paper.

So $V = (V' + 0.018y + 0.24)$ where y is the length (cm) of the capillary above the mercury half-way through the test.

The "GO" permeability coefficients of poly(4-methyl pent-1-ene) and polystyrene determined with this apparatus under oxygen-to-vacuum conditions at 20°C, 25°C and 34°C are shown in Table IV.1 (see Appendix 4).

As shown in the table, both the "GO" permeability coefficients of poly(4-methyl pent-1-ene) and polystyrene increased with temperature, and this could be explained as being due to an increase in the flexibility

of the polymer chain and hence an increase in the free volume in the polymer as the temperature was increased. The "GO" permeability coefficient of poly(4-methyl pent-1-ene) obtained at 25°C by this method (226×10^{-10} units) was lower than that given by the Metal Box Ltd.[213] (275×10^{-10} units), (where "units" here and hereafter means cc(S.T.P.). $mm.cm^{-2}.sec^{-1}.cm\ Hg^{-1}$), possibly due to the different apparatus and techniques used in both works, because the poly(4-methyl pent-1-ene) sample used in this work was obtained from the Metal Box Ltd.

The Davenport Gas Permeability Measurement techniques required the pressure reading to be taken manually, and several errors could possibly arise from this method, such as (a) the difficulty in reading accurately, to the second decimal point in cm Hg, the mercury level in the capillary, (b) false reading due to the presence of bubbles trapped in the mercury column during the course of the experiment, and (c) leak from the stop-cock of the apparatus, or from the permeation cell holding the polymer film.

The Davenport Gas Permeability Apparatus was not applicable to the measurement of oxygen permeability in contact lens applications in the sense this technique measured only the "GO" transmission of a polymer film, and could not measure its "DO" permeability which is of greater significance in the contact lens field. However the Davenport Gas Permeability Apparatus helped to check the validity of the application of a membraned electrode to oxygen permeability measurements in this work, which will be discussed later in this chapter.

IV.3.A.2. The Modified E.I.L. Electrode System

The idea of using a polarographic oxygen sensor in the oxygen permeability studies was that it was hoped it could measure both

"gaseous" oxygen and "dissolved" oxygen. In the experiments involving the E.I.L. electrode system, only "GO" permeabilities of some hydrophobic materials were first studied to see the validity of this piece of apparatus in the oxygen permeability measurements.

During calibration of this apparatus, the millivolt shown on the electrometer was found to vary quite linearly with the oxygen pressure between zero and 76 cm Hg. Four hydrophobic materials were used in the oxygen permeability measurements under oxygen-to-air conditions: poly-(4-methyl pent-1-ene), ethylene propylene terpolymer, silicone rubber and oriented polypropylene (the first three being potential contact lens materials). From the oxygen pressure trace on the recorder sheet in each case, a plot of $\ln(76-p_2)$ vs. t (in mins.) gave a slope k (min^{-1}), and then the "gaseous" oxygen permeability coefficient was calculated from Equation III.10a:

$$P_g = \frac{V_2}{A} \times \frac{1}{76} \times \frac{273}{273+T} \times \frac{kL}{60} \qquad \text{(IV.10a)}$$

Several attempts were made in choosing a gasket for making the volume (V_2) of the oxygen-receiving side (i.e. side 2) of the permeation cell, and in this respect, a commercial rubber "O" ring (internal diameter A=0.16 cm^2), a flat metal washer (A=0.196 cm^2) and a "home-made" flat rubber "O" ring (A=0.238 cm^2) were tried, and the results on the permeability measurements are shown in Table IV.2 (see Appendix 4).

Comparisons of the results shown in Table IV.2 and Table IV.3 (literature values) show that the P_g values could be different as much as two fold or more (although the small difference in temperature may have a small effect on the "GO" permeability coefficient), and that there is no correlation between the two sets of the results. Thus this

apparatus using the E.I.L. electrode system was not satisfactory even in the "GO" permeability measurements.

No attempt was made further to modify this apparatus (with the E.I.L. electrode system) because in addition to the erroneous results mentioned above, some disadvantages associated with this method were found during the course of the work: (a) the electrode system was found to be sometimes unstable, and also to become less sensitive easily, (b) it was difficult to know accurately the volume (V_2) of the side 2 of the permeation cell, and also it was rather impossible technically to introduce a nitrogen-saturated water into the volume V_2 of the permeation cell if it were to be used in "dissolved" oxygen permeability measurements under water/water conditions, and (c) some polymer samples,particularly soft material such as silicone rubber, were found to be easily ruptured due to the shearing force during the tightening of the top half of the permeation cell on to the sample placed on the bottom half of the cell.

IV.3.A.3. The I.M.I.Oxygen Sensor and the Large Permeation Cell

Both the "gaseous" and "dissolved" permeability coefficients of some potential contact lens materials were measured using this piece of apparatus. The apparatus was calibrated to read zero in pure nitrogen and full-scale deflection in pure oxygen, and the monitor reading was found to vary linearly with the oxygen pressure (with the scale reading at 21% saturation in ambient air). In water, the sensitivity of the gas-calibrated oxygen sensor was found to decrease accordingly, and the extent depended on whether the water was stirred or unstirred, as shown in Table IV.4 (see Appendix 4). However, owing to ease and quickness, calibration of the oxygen sensor was always made in gas phase.

The "gaseous" oxygen ("GO") permeability coefficients of poly(4-methyl pent-1-ene) and ethylene propylene terpolymer under gas/gas conditions were calculated from Equation IV.10a, and the results are shown

$$P_g = \frac{V_2}{A} \times \frac{1}{76} \times \frac{273}{273+T} \times \frac{kL}{60} \qquad \text{(IV.10a)}$$

(where k in min^{-1})

in Table IV.5 (see Appendix 4). The "GO" permeability coefficient of poly(4-methyl pent-1-ene) measured at 22°C by this method was found to be quite close to that at 25°C given by the Metal Box Ltd. (compare Tables IV.5 and IV.3).

The "dissolved" oxygen ("DO") permeability coefficients, P_d, of poly(4-methyl pent-1-ene) and HEMA-St (90:10) hydrogel at 34°C were calculated from Equation IV.17a, and the results are shown in Table IV.6 (see Appendix 4).

$$P_d = \frac{V_2 \cdot S_W}{A} \times \frac{kL}{60} \qquad \text{(IV.17a)}$$

(where k in min^{-1})

For poly(4-methyl pent-1-ene), the "DO" permeability coefficient was lower than the "GO" permeability coefficient, (compare Tables IV.6 and IV.5), and this could be explained by the resistance to the oxygen permeability due to the boundary layer between water and the polymer sample under the water/water conditions. For soft materials such as HEMA-St (90:10) hydrogel, the volume V_2 of the side 2 of the permeation cell was found to increase to about twice the volume in the case of poly(4-methyl pent-1-ene) due to sinking of the soft material during the clamping step, and the volume increase was taken into consideration in the calculation of P_d value.

Although the I.M.I. oxygen sensor used here behaved quite well in the oxygen permeability measurements (both gas/gas and water/water cases),

some disadvantages were found to be associated with the use of the large permeation cell, e.g. (a) a large test sample was required (i.e. a flat disc of about 7.6 cm in diameter), (b) as a large test sample was used, a soft and weak hydrogel material was more likely to be ruptured owing to the weight of water in the top half (i.e. side 2) of the permeation cell acting on the sample, and this was experimentally encountered and (c) it was difficult to make a large polymer sample of uniform thickness.

Therefore a small permeation cell was designed and made to eliminate these problems, and will be discussed in the following section.

IV.3.A.4. <u>The I.M.I. Oxygen Sensor and the small Permeation Cell</u>

Using this piece of apparatus, a great deal of work was carried out on both the "GO" and "DO" permeabilities of many materials including hydrogels because all the results obtained (see later in Section IV.3.B) indicate that this was the most satisfactory among all the methods that had been attempted in this work.

The results on the calibration of the I.M.I. oxygen sensor were found to be the same as those mentioned in Section IV.3.A.3, and the "GO" and "DO" permeability coefficients were calculated using Equations IV.10a and IV.17a respectively.

Four methods were attempted in the mounting of a polymer sample in the permeation cell in order that a good seal could be obtained and at the same time the polymer sample remained intact. The sample mounting methods are described as follows:

<u>Method 1</u>: The polymer sample (2.1 cm diameter, cut with the Griffins and George's cork-borer, size 14) was placed on the bottom "O" ring R_B (see Figure IV.7), and then the top "O" ring R_A set on it. Fragile hydrogels were found to be easily ruptured by this mounting technique,

but tough materials such as poly(4-methyl pent-1-ene), polystyrene and low water content hydrogels behaved well.

Method 2: The polymer sample (1.8 cm in diameter) was placed on the flange of the bottom chamber of the permeation cell (F_B in Figure IV.7) and the flange (F_A) of the top chamber sat on the sample while the rubber "O" rings sat on each other. If the sample was not thick enough (say, 0.35 mm) to ensure a good contact between the top flange, the sample and the bottom flange, leaks might occur.

Method 3: In this method, the hydrogel test sample (1.8 or 1.95 cm diameter) was placed on a poly(4-methyl pent-1-ene) sample (2.1 cm diam.) ensuring a good contact between the two. With the hydrogel sample facing the oxygen probe, the "laminate" was placed on the rubber "O" ring of the bottom half of the permeation cell so that the excess 0.3 cm or 0.15 cm width on the edge of the poly(4-methyl pent-1-ene) sheet was clamped between rubber "O" rings without rupturing the fragile hydrogel. Alternatively, both the poly(4-methyl pent-1-ene) and hydrogel samples were 1.8 cm in diameter, and the "laminate" was placed between the flanges of the permeation cell, the poly(4-methyl pent-1-ene) sheet again acting as a support to the soft hydrogel.

Method 4: This method was developed concurrently with Method 3. For thin samples of hydrogels (about 0.2 mm thick), it was found from many measurements that a good seal was achieved by spreading a layer of plasticine on both the top and bottom flanges of the permeation cell, and covering part of the plasticine and the rubber "O" ring (R_B) of the bottom chamber with a thin polythene "O" ring (about 0.1 mm thick), and then placing the test sample (1.8 cm diameter) on the polythene "O" ring and the remainder of the plasticine. (See Figure IV.9). The use of the polythene "O" ring was later found to be optional if a correct amount of plasticine was applied to both the top and bottom flanges, and this could

be learnt from experience. However, this sample mounting method was used in most measurements for hydrogels especially those having high water content.

Besides the problem of sample mounting, another obstacle was encountered during the development of the oxygen permeability measurements under water/water conditions. A negative oxygen monitor reading was observed in some cases when water was started to pump into the bottom chamber of the permeation cell (when the oxygen permeability was to start) apparently due to the disturbance of the electrolyte in the oxygen sensor. This was found to be eliminated if water was pumped into an intermediate container from which water was then allowed, due to hydrostatic pressure, to run through the bottom chamber of the permeation cell at a rate of about three drops per second. So most of the measurements were carried out with this set-up.

The results of the oxygen permeability coefficients of some gels and hydrophobic materials will be shown and discussed in the following section (Section IV.3.B).

Several difficulties were found to be associated with the oxygen permeability measurements using this method and affect consequently the accuracy of the results:

(a) There might be leaks at the edge of the polymer sample, but this can be detected during the flushing of the top chamber of the permeation cell with a nitrogen gas when the bottom chamber was filled with a nitrogen-saturated water. Use of a sufficient amount of plasticine as the seal and a uniform application of the clamping pressure on the sample were both found to reduce the leak.

(b) There might be rupture of the sample, especially with the high water content hydrogels. But this again could be detected and the chance of rupture was very much reduced in this case where only a small

quantity of water was used in the top chamber of the permeation cell. Also avoidance of contact between the soft sample and the syringe needle during flushing the top chamber with the nitrogen gas and filling with the nitrogen-saturated water helped to reduce the chance of sample rupture.

(c) Due to the small volume (less than 1 cc) of the top chamber (i.e. side 2) and the small size of the sample, uncertainty in any of them (particularly the former) resulted in a big discrepancy in the oxygen permeability coefficient. This error might be reduced by measuring accurately the quantity of the nitrogen-saturated water being introduced into that chamber with a small syringe. However the uncertainty in the oxygen permeability coefficient arising from this volume factor could be as high as 10%.

(d) Any bubble formed at the trough near the electrode gave erroneous results. The bubble was more likely to be there during the introduction of the nitrogen-saturated water, and this could be avoided by inverting the permeation cell at an angle and slowly filling the top chamber while tapping the cell and viewing through the transparent bottom of the cell.

(e) Disturbance of the electrolyte in the oxygen sensor also gave erratic results. This should not happen in the absence of bubbles in the electrolyte, electrical disturbance or accidental unscrewing of the cap of the oxygen sensor.

(f) The plot of $\ln(p_1 - p_2)$ vs. t was found in some measurements under water/water conditions to curve upwards but linearitywas still observed in the first 50% (or more) of the plot. This could be due to insufficient mixing of oxygen and the water in the top chamber of

the permeation cell (because it was unstirred), and/or to the high oxygen consumption rate of the oxygen sensor, and/or concentration dependence of the oxygen permeability coefficient. However, the slope (k) of the straight line portion of the graph in all cases was taken between the first 10% and 20% of the graph (similar to the contact lens case where the critical oxygen tension (about 15 mm Hg) under the lens was about 10% of the atmosphere oxygen (159 mm Hg)). In those cases where it was difficult to draw a straight line in this region, the measurements were discarded. It has been suggested[214] that oxygen permeability coefficient value obtained in an unstirred solution was lower than in vigorously stirred conditions, but this observation related to permeation cells having a relatively large receiving volume (i.e. side 2). Due to the small volume (< 1 ml) of the oxygen receiving side of our permeation cell, it was difficult to stir the water in that chamber, and so oxygen permeability measurements were carried out under unstirred conditions in this work (although the air-saturated water in the bottom chamber was vigorously stirred). It seems that the need to stir the receiving cell disappeared as the volume became smaller. The consistency of the various measurements described later will confirm this.

IV.3.B. OXYGEN PERMEABILITY COEFFICIENTS OF SOME POTENTIAL CONTACT LENS MATERIALS

Using the I.M.I. oxygen sensor and the small permeation cell, the oxygen permeability coefficients of many materials were measured, and the results could be classified into three groups: (1) hydrophobic materials, (2) "laminates" of hydrogels and poly(4-methyl pent-1-ene), and (3) hydrogel materials (and hydrophilic materials which are non-gels on hydration.

IV.3.B.1. Hydrophobic Materials

The "gaseous" oxygen ("GO") permeability coefficients of poly(4-methyl pent-1-ene) and polystyrene measured at 22°C are shown in Table IV.7 (see Appendix 4) and agreed quite well with those values measured with the Davenport Gas Permeability Apparatus (Table IV.1). The dissolved oxygen ("DO") permeability coefficients of the same materials measured at 25°C and 34°C (at various thicknesses) under water/water conditions are shown in Table IV.8 (see Appendix 4).

The "DO" permeability coefficient of poly(4-methyl pent-1-ene) was found to be lower than its "GO" permeability coefficient at the same temperature (Tables IV.7 and IV.8). Also at a given temperature, the "DO" permeability coefficient increased with the sample thickness (Table IV.8). These observations could be explained by the additional resistance to oxygen permeation at the boundary between water and the polymer film. In the oxygen permeability measurements under gas/gas conditions, the boundary effect was negligible due to quick equilibrium between the gas and the polymer sample, but in the water/water conditions, the boundary effect had to be taken into consideration. The equation which has been derived[215] to describe "DO" permeability of silicone

rubber with the consideration of the boundary effect is given by Equation IV.22.

$$\frac{1}{P_d} = \frac{1}{DS} + R.\frac{1}{L} \qquad \text{(IV.22)}$$

where P_d is the "DO" permeability coefficient of the sample at that thickness;

D and S are the diffusion coefficient and solubility of the sample respectively

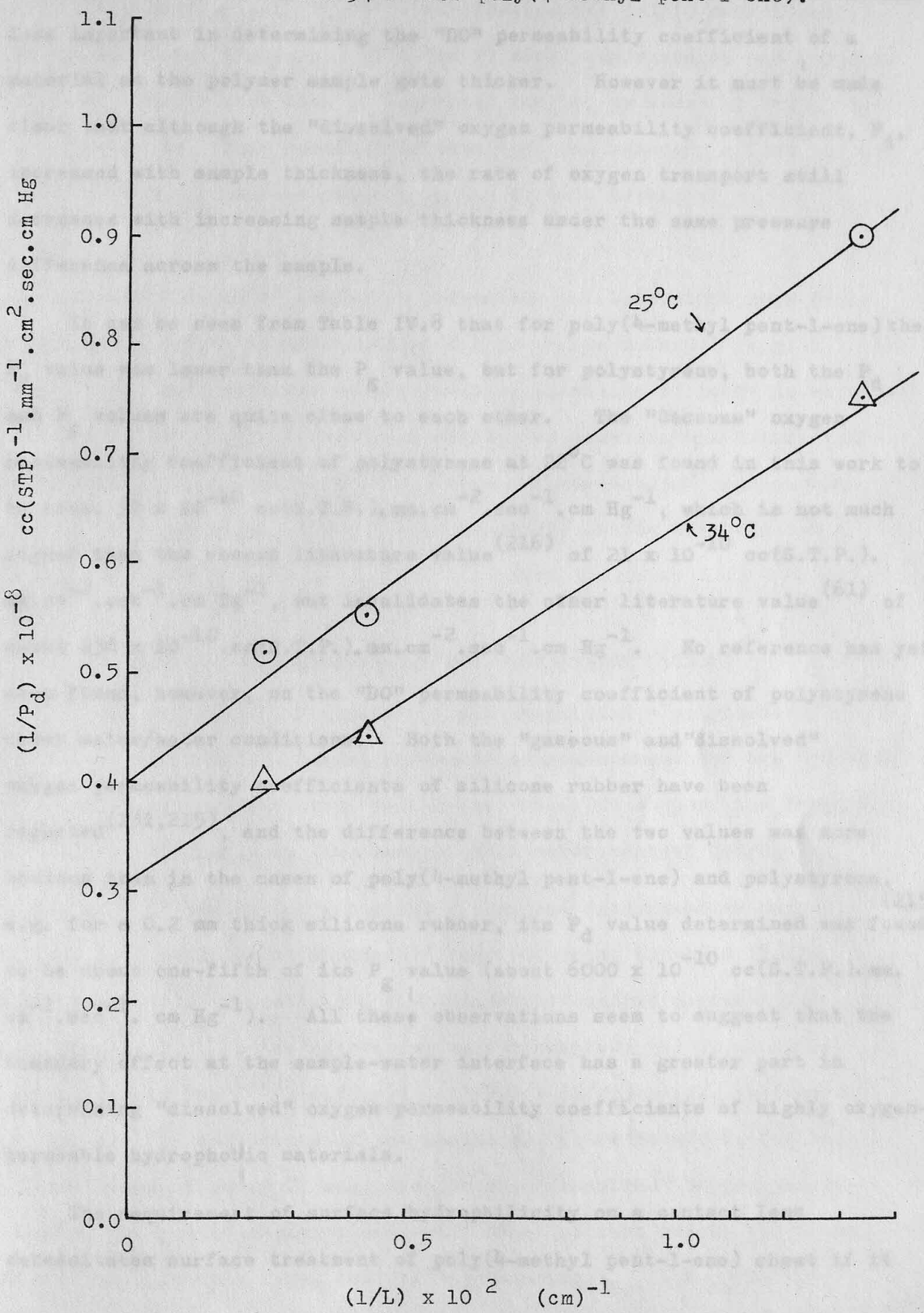
R is the resistance to the oxygen permeability due to the boundary effect

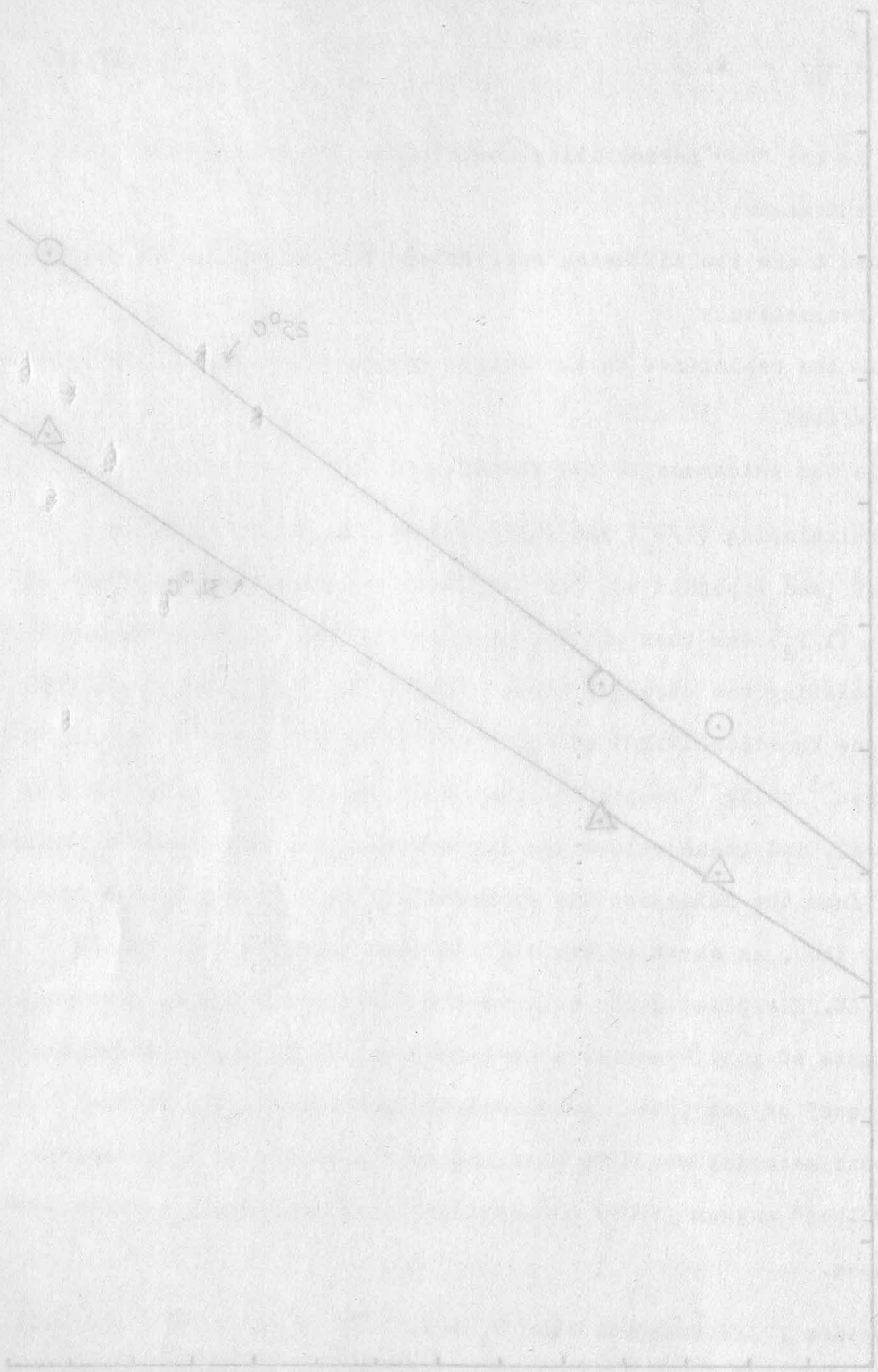
and L is the thickness of the sample.

By calculating $(1/P_d)$ and $(1/L)$ values, which are shown in Table IV.9 (see Appendix 4), for poly(4-methyl pent-1-ene) at both 25°C and 34°C, $(1/P_d)$ was then plotted against $(1/L)$ as shown in Figure IV.10. By extrapolating the straight lines back to the $(1/P_d)$ axis gave (DS) values (see Equation IV.22) of 250 x 10^{-10} and 322 x 10^{-10} cc (S.T.P.). $mm.cm^{-2}.sec^{-1}.cm\ Hg^{-1}$ respectively at 25°C and 34°C for poly(4-methyl pent-1-ene), and these values are quite comparable with those P_g values obtained from the Davenport Gas Permeability Apparatus and also from the Metal Box Ltd., as shown in Table IV.10 (see Appendix 4). Hence Equation IV.22 applied quite well to the "dissolved" oxygen permeability measurements of poly(4-methyl pent-1-ene) and in general, therefore, the "gaseous" oxygen ("GO") permeability coefficient, P_g, of a hydrophobic material could be obtained from a series of measurements of "dissolved" oxygen ("DO") permeability coefficients, P_d, at various thicknesses.

Equation IV.22 suggests that $P_d = D.S = P_g$ when $L = \infty$, and this means that the P_d value of a material increases and approaches its P_g value as the sample thickness is increased, and becomes equal to P_g when sample

Figure IV.10 The plot of $(1/P_d)$ vs. $(1/L)$ at 25°C and and 34°C for poly(4-methyl pent-1-ene).





is infinitely thick. In other words, the boundary effect observed in the oxygen permeability measurements under water/water conditions becomes less important in determining the "DO" permeability coefficient of a material as the polymer sample gets thicker. However it must be made clear that although the "dissolved" oxygen permeability coefficient, P_d, increased with sample thickness, the rate of oxygen transport still decreases with increasing sample thickness under the same pressure difference across the sample.

It can be seem from Table IV.8 that for poly(4-methyl pent-1-ene) the P_d value was lower than the P_g value, but for polystyrene, both the P_d and P_g values are quite close to each other. The "Gaseous" oxygen permeability coefficient of polystyrene at 22°C was found in this work to be about 37 x 10^{-10} cc(S.T.P.).mm.cm^{-2}.sec^{-1}.cm Hg^{-1}, which is not much higher than the recent literature value[216] of 21 x 10^{-10} cc(S.T.P.). mm.cm^{-2}.sec^{-1}.cm Hg^{-1}, but invalidates the other literature value[61] of about 250 x 10^{-10} cc(S.T.P.).mm.cm^{-2}.sec^{-1}.cm Hg^{-1}. No reference has yet been found, however, on the "DO" permeability coefficient of polystyrene under water/water conditions. Both the "gaseous" and"dissolved" oxygen permeability coefficients of silicone rubber have been reported[131,215], and the difference between the two values was more obvious than in the cases of poly(4-methyl pent-1-ene) and polystyrene, e.g. for a 0.2 mm thick silicone rubber, its P_d value determined was found[215] to be about one-fifth of its P_g value (about 6000 x 10^{-10} cc(S.T.P.).mm. cm^{-2}.sec^{-1}. cm Hg^{-1}). All these observations seem to suggest that the boundary effect at the sample-water interface has a greater part in determining "dissolved" oxygen permeability coefficients of highly oxygen-permeable hydrophobic materials.

The requirement of surface hydrophilicity on a contact lens necessitates surface treatment of poly(4-methyl pent-1-ene) sheet if it

were to be used in contact lens applications. The "dissolved" oxygen permeability coefficient, P_d, of a surface-treated poly(4-methyl pent-1-ene) (chromic acid treatment: 60°C for 15 mins) was measured and found to be slightly higher than that of untreated sample, as shown in Table IV.11 (see Appendix 4). This result indicates that the boundary effect in the oxygen permeation had been reduced by the surface treatment of the polymer.

The significance of the above discussion in the contact lens field is that since oxygen permeability of a hydrophobic material (e.g. silicone rubber) drops and decreases with decreasing thickness if water is at both sides of the material, it is more realistic to consider the "dissolved" oxygen permeability coefficient, P_d, of a surface-treated hydrophobic material at a particular contact lens thickness in the selection of a contact lens material.

IV.3.B.2. "Laminate" of Hydrogels and Poly(4-methyl pent-1-ene)

The idea of using a "laminate" of a hydrogel and a poly(4-methyl pent-1-ene) sheet in the oxygen permeability measurement of the hydrogel was to allow the poly(4-methyl pent-1-ene) sheet to support the fragile hydrogel (particularly in the cases of high water content hydrogels).

The k_{Lam} of the ln $(p_1 - p_2)$ vs. t plot was determined as before, and thus P_{Lam} could be determined if required using Equation IV.17a, (Lam = laminate). To calculate the "dissolved" oxygen permeability coefficient, $(P_d)_{Gel}$, two approaches were attempted: (i) by using the "laminate" equation[217], $(1/P_{Lam})$ $(1/P_1) + (1/P_2) + ...$(where 1,2, ... are the components of the laminate), to calculate P_2 for the hydrogel in question if P_1 was taken as the "dissolved" oxygen permeability of the poly(4-methyl pent-1-ene) sheet at that sample thickness, and was known in this case, and (ii) by assuming $k_{Gel} = k_{Lam}$ to calculate $(P_d)_{Gel}$ using Equation IV.17a. The approach (i) was found

not applicable to the system here because the $(P_d)_{Gel}$ values thus obtained (not shown in Table IV.12) were higher than the values determined in the cases with no poly(4-methyl pent-1-ene) sheet as the support.

The $(P_d)_{Gel}$ values calculated using the approach (ii) were found to be quite close to the values obtained in the cases with no poly(4-methyl pent-1-ene) supporting sheet, if the poly(4-methyl pent-1-ene) sheet was sufficiently thin not to affect the oxygen permeability of the hydrogel in question. As shown in Table IV.12 (see Appendix 4), for examples in the cases of PHEMA and HEMA-VP (50:50) hydrogels, if the poly(4-methyl pent-1-ene) supporting sheet was 0.23 mm. thick or more, the $(P_d)_{Gel}$ values calculated by assuming $k_{Gel} = k_{Lam}$ (i.e. approach (ii)) were found to be lower than the values obtained in the cases with only the hydrogel on its own, but if the poly(4-methyl pent-1-ene) sheet was 0.075 mm. thick, then the two values determined by the two techniques became quite consistent. On the other hand, with low-oxygen-permeable hydrogels such as HEMA-St (90:10) hydrogel, a 0.23 mm. thick poly(4-methyl pent-1-ene) supporting sheet was found to have a little effect on the "dissolved" oxygen permeability coefficient, P_d, of the hydrogel thus determined, as in the case of a 0.075 mm. thick poly(4-methyl pent-1-ene) support (see Table IV.12).

In almost all the cases studied (i.e. those with a sufficiently thin poly(4-methyl pent-1-ene) sheet as a support), as shown in Table IV.12 the results seemed to indicate that the "laminate" method could be used to determine the "dissolved oxygen permeability coefficient of a hydrogel provided that (i) $k_{Gel} = k_{Laminate}$ and used in the calculation of $(P_d)_{Gel}$ and (ii) the sheet supporting the hydrogel in the measurement must have such a higher rate of oxygen transmission (i.e. highly oxygen permeable and sufficiently thin) than the hydrogel that it has little effect on the

oxygen permeability of the hydrogel as in the case if it were not there. A good contact between the hydrogel and the supporting sheet seemed to be required in this "laminate" to obtain reproducible results. With low water content hydrogels, the type of contact just mentioned might not be good, but these hydrogels were tough enough that their oxygen permeabilities could be measured on their own without any supporting sheet. With high water content hydrogels which were soft and fragile, the contact between any of these hydrogels and the poly(4-methyl pent-1-ene) was good, and so the "dissolved" oxygen permeability coefficient of the hydrogel could be determined using this "laminate" method. Although this "laminate" method requires "dissolved" oxygen permeability coefficients of hydrogels to be measured independently for comparison, all the experimental results obtained so far suggest that the "laminate" method is applicable to the determination of the "dissolved" oxygen permeability coefficient of a hydrogel provided all the requirements mentioned above are satisfied. This method is particularly useful if a hydro gel of diameter slightly smaller than 1.8 cm. is available or if a hydrogel is too fragile to be clamped on its own in the permeation cell.

IV.3.B.3. Hydrogel Materials

As mentioned eariler, an application of an appropriate quantity of plasticine to give a good seal in the permeation cell enables "dissolved" oxygen permeability coefficients of hydrogels to be measured on their own without any support.

The flow rate of the air-saturated water running continuously through the bottom half of the permeation cell was found to have little effect on the "dissolved" oxygen permeability coefficient of PHEMA hydrogel measured, as shown in Table IV.13 (see Appendix 4). However most

measurements were carried out at a water flow rate of about two or three water drops per second.

The effect of sample thickness on the "dissolved" oxygen permeability coefficient, P_d, of PHEMA hydrogel was also investigated at 25°C and 34°C. Two PHEMA hydrogels of different thicknesses (0.21 and 0.39 mm.) but similar equilibrium water content were used, and the results are shown in Table IV.14 (see Appendix 4). Due to the limitation of the small permeation cell which could not hold any hydrogel material of thickness more than 0.5 mm., it was difficult to see if the "dissolved" oxygen permeability coefficient of PHEMA hydrogel was constant over a wide range of sample thicknesses. However, the "dissolved" oxygen permeability coefficient of a hydrogel material was not affected by its sample thickness at least between about 0.2 mm. and 0.4 mm. (as shown in the case of PHEMA hydrogel), which is the range of thickness currently used for some contact lenses.

The results on the "dissolved" oxygen permeability coefficients of many hydrogels of different chemical structures and equilibrium water contents at 25°C and 34°C (normal eye temperature) are shown in Table IV. 15, and Table IV.16 respectively (see Appendix 4).

The "dissolved" oxygen ("DO") permeability coefficient of PHEMA hydrogel at 40% equilibrium water content was found in this work to be about 75 x 10^{-10} cc (S.T.P.).mm.cm^{-2}.sec^{-1}.cm Hg^{-1} at 25°C, which is somewhere between the values of 51 x 10^{-10} cc(S.T.P.).mm.cm^{-2}.sec^{-1}.cm Hg^{-1} reported by Yasuda et al[131,134] and 89 x 10^{-10} cc(S.T.P.)mm.cm^{-2}.sec^{-1}. cm Hg^{-1} reported by Fatt and St.Helen[136]. These discrepancies could possibly be due to different experimental techniques and measuring apparatus used in the various works. However it can be concluded from the present work that the other value of 4 x 10^{-10} cc(S.T.P.)mm.cm^{-2}.sec^{-1}.

cm Hg^{-1} which has also been reported for PHEMA hydrogel by Takahashi et al[87], Kamath[135] and Fatt et al[61] and has caused disputes on the true "DO" permeability coefficient of PHEMA hydrogel is certainly wrong because in their works an oxygen gas was passed through the hydrogel sample during the oxygen permeability measurement and apparently this dried up the sample, giving a false "DO" permeability coefficient of PHEMA hydrogel and instead giving effectively the "gaseous oxygen permeability coefficient of a dehydrated PHEMA.

In the present work in which only one side (i.e. side 1) of the hydrogel test sample was stirred during measurements, the "DO" permeability coefficient of PHEMA hydrogel at 25°C was found to be quite close to the value obtained in the work by Yasuda et al[134], in which both sides of the test specimen were stirred vigorously during measurements. This result indicates that the necessity to stir the water on the side 2 of a hydrogel test sample in the oxygen permeability measurement diminished as the volume of that chamber (i.e. side 2) became smaller, such as the one used in the present work.

The "DO" permeability coefficient of PHEMA hydrogel (E.W.C. 40%) at 34°C was determined in this work as about 145 x 10^{-10} cc(S.T.P.)mm. $cm^{-2}.sec^{-1}.cm\ Hg^{-1}$ cf. 110x10^{-10} obtained by Fatt& St.Helen on Soflens lenses (also PHEMA hydrogel) of 39% E.W.C. This discrepancy could be due to the partial drying-up of the test sample in Fatt and St.Helen's work, who used an infra-red lamp to maintain the test temperature at 34°C during the oxygen measurement. No other literature work has yet been found on the "dissolved" oxygen permeability coefficient of PHEMA at 34°C.

The "dissolved" oxygen (i.e. "DO") permeability coefficients, P_d, of hydrated HEMA-St copolymers were plotted against HEMA content in Figure IV.11 and against their equilibrium water contents in Figure IV.12.

The P_d value decreased as the HEMA content in the copolymer was increased from none to about 50%, and then started to reverse and increased markedly at about 75% HEMA content. The early part of the curve suggests that dehydrated PHEMA was less oxygen-permeable than polystyrene because increasing HEMA content in the copolymer reduced the free volume and hence the oxygen permeability of the copolymer, and that water which was thought to bind to the polar groups of the copolymers in the hydrated state in that region of copolymer compositions seemed to have no noticeable effect on the oxygen permeability. The marked increase in the "dissolved" oxygen permeability coefficient in the latter part of the curve corresponded quite well to the reversal of the density-copolymer composition graph (Figure II.8) and also to the marked increase in the equilibrium water contents of the hydrogels in that region of copolymer compositions (shown earlier in Figure III.1), and this perhaps could be attributed to an increase in the free volume of the hydrogels for oxygen permeability with increasing equilibrium water content. Liquid water has been reported[214] to have a "dissolved" oxygen permeability coefficient, P_d, of 930 x 10^{-10} cc(S.T.P.).mm.cm^{-2}.sec^{-1}. cm Hg^{-1} at 25°C, which is several times greater than the P_d values of all these HEMA-St copolymer hydrogels, and so water is expected to have a larger free volume, as also shown in its very low glass transition temperature of -137°C[218].

By and large, it is very likely that for any homopolymer (e.g. poly-(4-methyl pent-1-ene)) having a higher "gaseous" oxygen permeability coefficient, P_g, than polystyrene, the "dissolved" oxygen permeability coefficients of its copolymers with HEMA (or any other hydrophilic monomer) in the hydrated state will have a similar correlation with copolymer compositions as in the case of hydrated HEMA-St copolymers, but perhaps with different curvatures and different reversing points. On the other

Figure IV.11 The correlations between the "dissolved" oxygen permeability coefficients of hydrated HEMA-St and HEMA-MMA copolymers at 25°C and 34°C with their copolymer compositions.

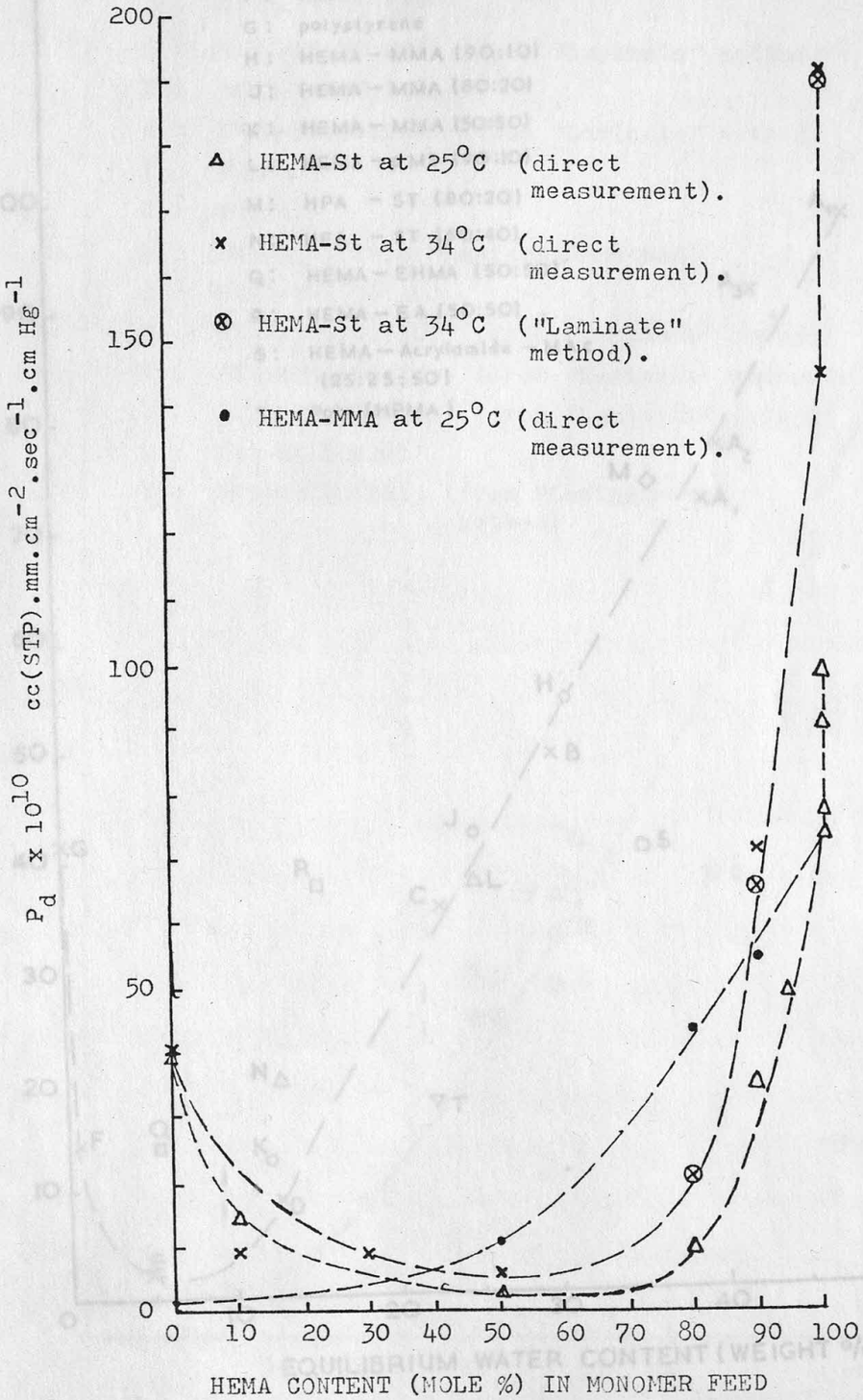


Figure IV.12 "Dissolved" oxygen permeability coefficients of various hydrated materials with different equilibrium water contents, at 25°C.

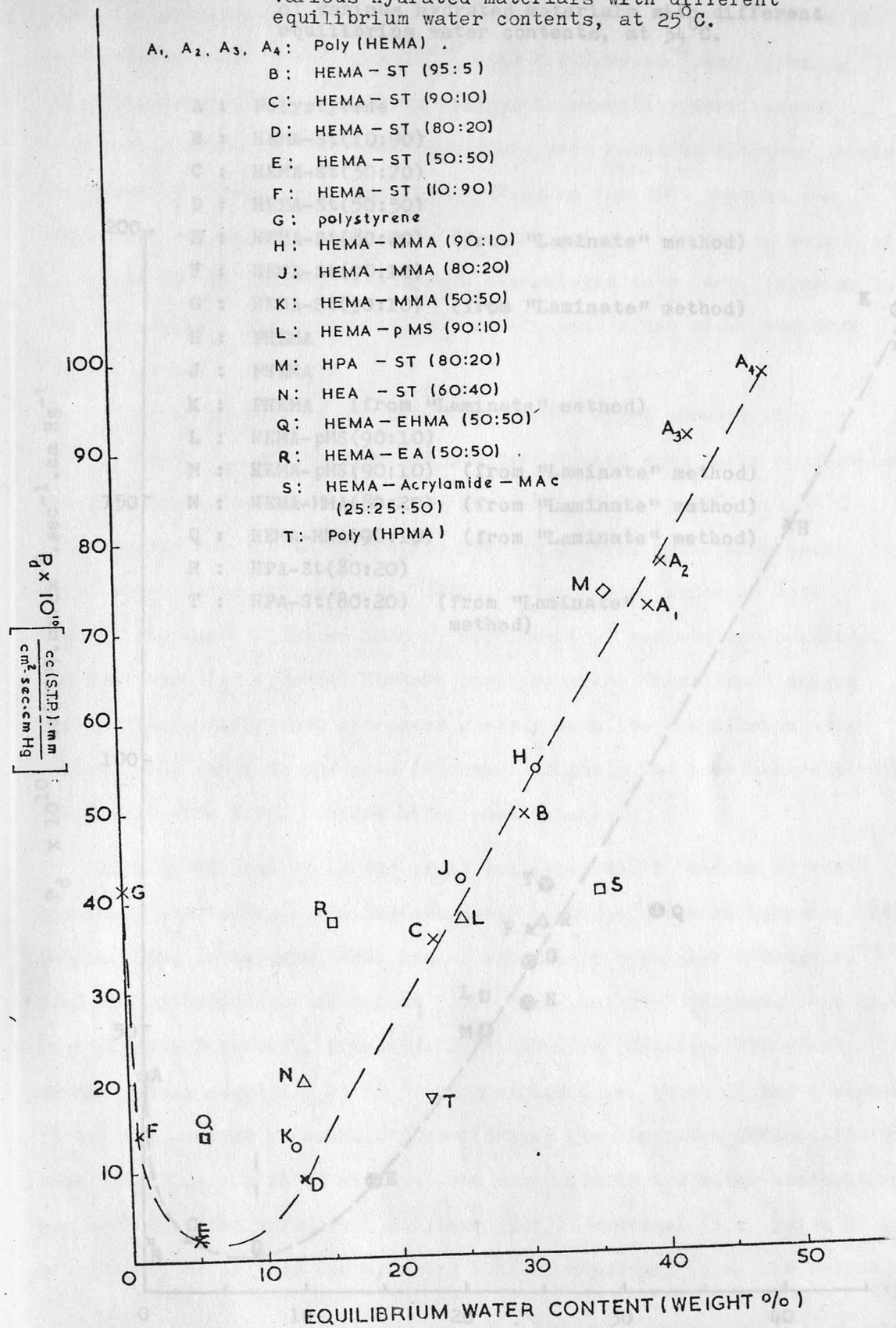
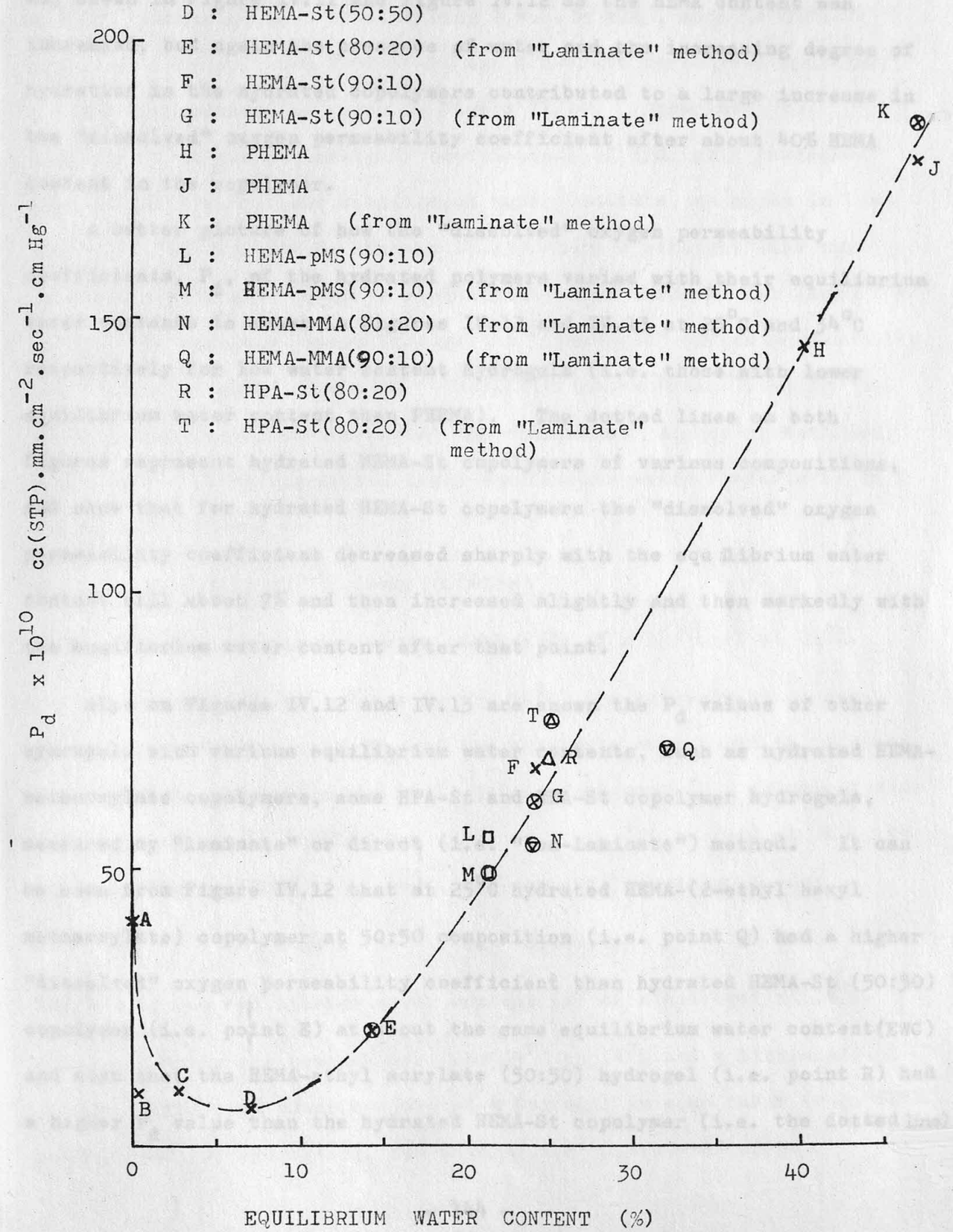


Figure IV.13 "Dissolved" oxygen permeability coefficients of various hydrated materials with different equilibrium water contents, at 34°C.

A : Polystyrene
B : HEMA-St(10:90)
C : HEMA-St(30:70)
D : HEMA-St(50:50)
E : HEMA-St(80:20) (from "Laminate" method)
F : HEMA-St(90:10)
G : HEMA-St(90:10) (from "Laminate" method)
H : PHEMA
J : PHEMA
K : PHEMA (from "Laminate" method)
L : HEMA-pMS(90:10)
M : HEMA-pMS(90:10) (from "Laminate" method)
N : HEMA-MMA(80:20) (from "Laminate" method)
Q : HEMA-MMA(90:10) (from "Laminate" method)
R : HPA-St(80:20)
T : HPA-St(80:20) (from "Laminate" method)



hand, for poly(methyl methacrylate) which has a lower "gaseous" oxygen permeability coefficient (2×10^{-10} cc(S.T.P.).mm.cm^{-2}.sec^{-1}. cm Hg^{-1}) than polystyrene, the "dissolved" oxygen permeability coefficients of hydrated HEMA-MMA copolymer copolymers were found to increase in the way shown in Figure IV.11 and Figure IV.12 as the HEMA content was increased, but again the presence of water and the increasing degree of hydration in the hydrated copolymers contributed to a large increase in the "dissolved" oxygen permeability coefficient after about 40% HEMA content in the copolymer.

A better picture of how the "dissolved" oxygen permeability coefficients, P_d, of the hydrated polymers varied with their equilibrium water contents is shown in Figures IV.12 and IV.13 at 25°C and 34°C respectively for low water content hydrogels (i.e. those with lower equilibrium water content than PHEMA). The dotted lines on both figures represent hydrated HEMA-St copolymers of various compositions, and show that for hydrated HEMA-St copolymers the "dissolved" oxygen permeability coefficient decreased sharply with the equilibrium water content till about 7% and then increased slightly and then markedly with the equilibrium water content after that point.

Also on Figures IV.12 and IV.13 are shown the P_d values of other hydrogels with various equilibrium water contents, such as hydrated HEMA-methacrylate copolymers, some HPA-St and HEA-St copolymer hydrogels, measured by "laminate" or direct (i.e. "non-laminate") method. It can be seen from Figure IV.12 that at 25°C hydrated HEMA-(2-ethyl hexyl methacrylate) copolymer at 50:50 composition (i.e. point Q) had a higher "dissolved" oxygen permeability coefficient than hydrated HEMA-St (50:50) copolymer (i.e. point E) at about the same equilibrium water content(EWC) and also that the HEMA-ethyl acrylate (50:50) hydrogel (i.e. point R) had a higher P_d value than the hydrated HEMA-St copolymer (i.e. the dotted line)

having similar E.W.C. This could be attributed by the greater flexibility or larger free volume of the copolymer hydrogels of both 2-ethyl hexyl methacrylate and ethyl acrylate than styrene.

By extending the correlation between P_d and E.W.C. to higher water content hydrogels (i.e. those having E.W.C. > 40%), such as HEMA-VP copolymer hydrogels and HEMA-VP-St and HPA-VP-St terpolymer hydrogels, a curve was obtained as shown in Figure IV.14. Apparently the "dissolved" oxygen permeability coefficients of hydrogels increased exponentially with their equilibrium water contents, as shown in the plot of $\ln(P_d)$ vs. E.W.C. (Figure IV.15) where a straight line was obtained over a wide range of equilibrium water contents (at least 20% to 80%) and the line passed through the "dissolved" oxygen permeability coefficient of water (930×10^{-10} cc(S.T.P.)mm.cm^{-2}.sec^{-1}. cm Hg^{-1}). An empirical equation correlating the "dissolved" oxygen permeability coefficients of hydrogels and their equilibrium water contents at 25°C can be thus deduced as

$$P_d = 14 \times 10^{-10} \cdot \exp(0.0418w)$$

where w is the equilibrium water content (%) of a hydrogel at 25°C.

On increasing the temperature to 34°C, the "dissolved" oxygen permeability coefficients, P_d, of the hydrogels were found to increase and a linear correlation was still obtained between (P_d) values of some hydrogels and their equilibrium water contents (20 to 70%) at 34°C, as shown in Figure IV.15. At 34°C, the empirical equation became

$$P_d = 24 \times 10^{-10} \cdot \exp(0.0443w)$$

where w is the equilibrium water content (%) of a hydrogel at 34°C.

Increasing the temperature from 25°C to 34°C had a little effect on the equilibrium water content of a hydrogel in some cases (e.g. PHEMA, HEMA-St (90:10) hydrogels), but doubled its "dissolved" oxygen

permeability coefficient. On the other hand, the equilibrium water contents of some hydrogels such as HEMA-VP (50:50) and HPA-VP-St (35:55;10) were reduced as the temperature was increased from 25°C to 34°C but their "dissolved" oxygen permeability coefficients at 34°C were about twice those of other hydrogels at 25°C having the same equilibrium water content. These results suggest that at a higher temperature, a hydrogel had a larger free volume for oxygen permeability, resulting mainly from an increase in free volume of water in the hydrogel.

In a recent publication[214], Holly and Refojo concluded from all previous works that the permeation of "dissolved" oxygen through hydrated contact lenses of acrylic and other hydrogels occurs in the water-filled interstices of the cross-linked polymer network. The results of the present work support this, and remark that the "dissolved" oxygen permeability of hydrated polymers (particularly those with low water contents) are affected by the chemical structure of the polymer as well.

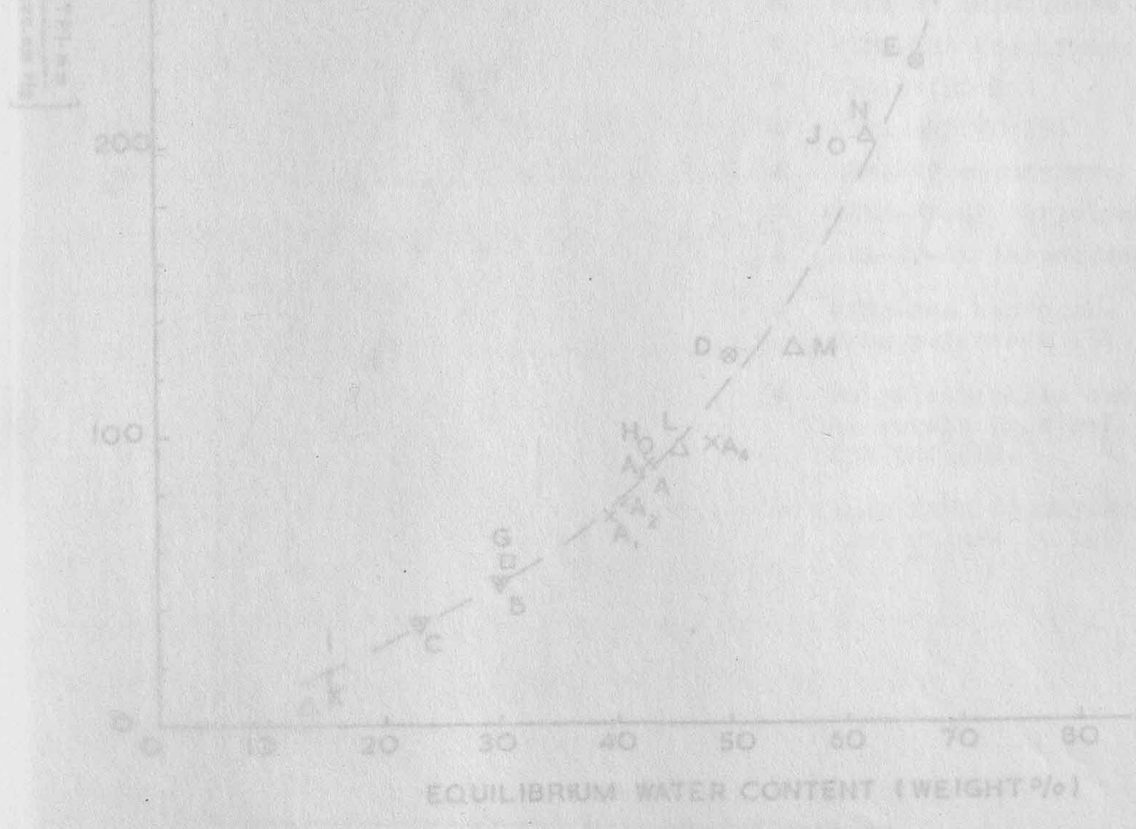


Figure IV.14 "Dissolved" oxygen permeability coefficients of various hydrated materials at 25°C.

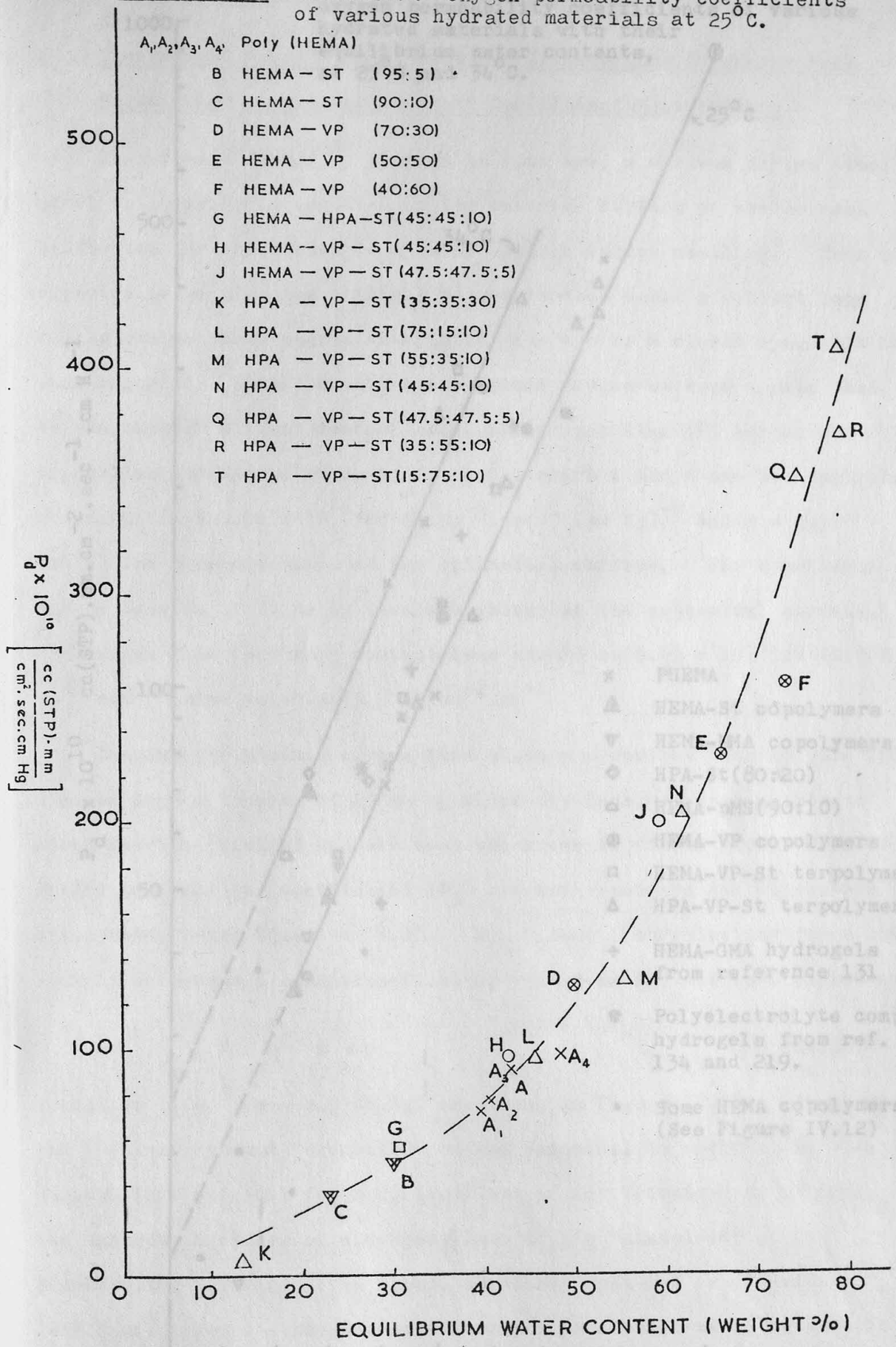
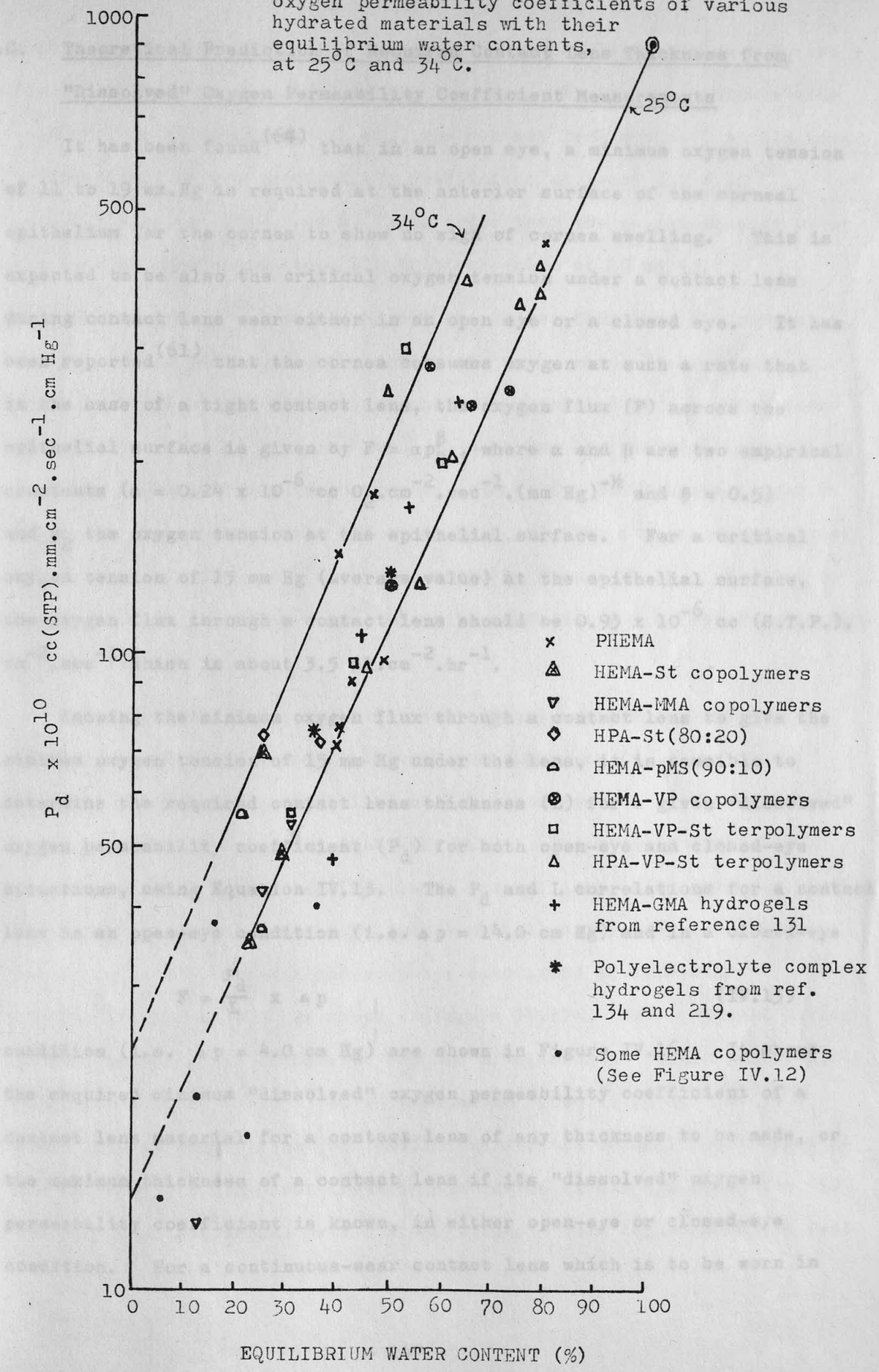
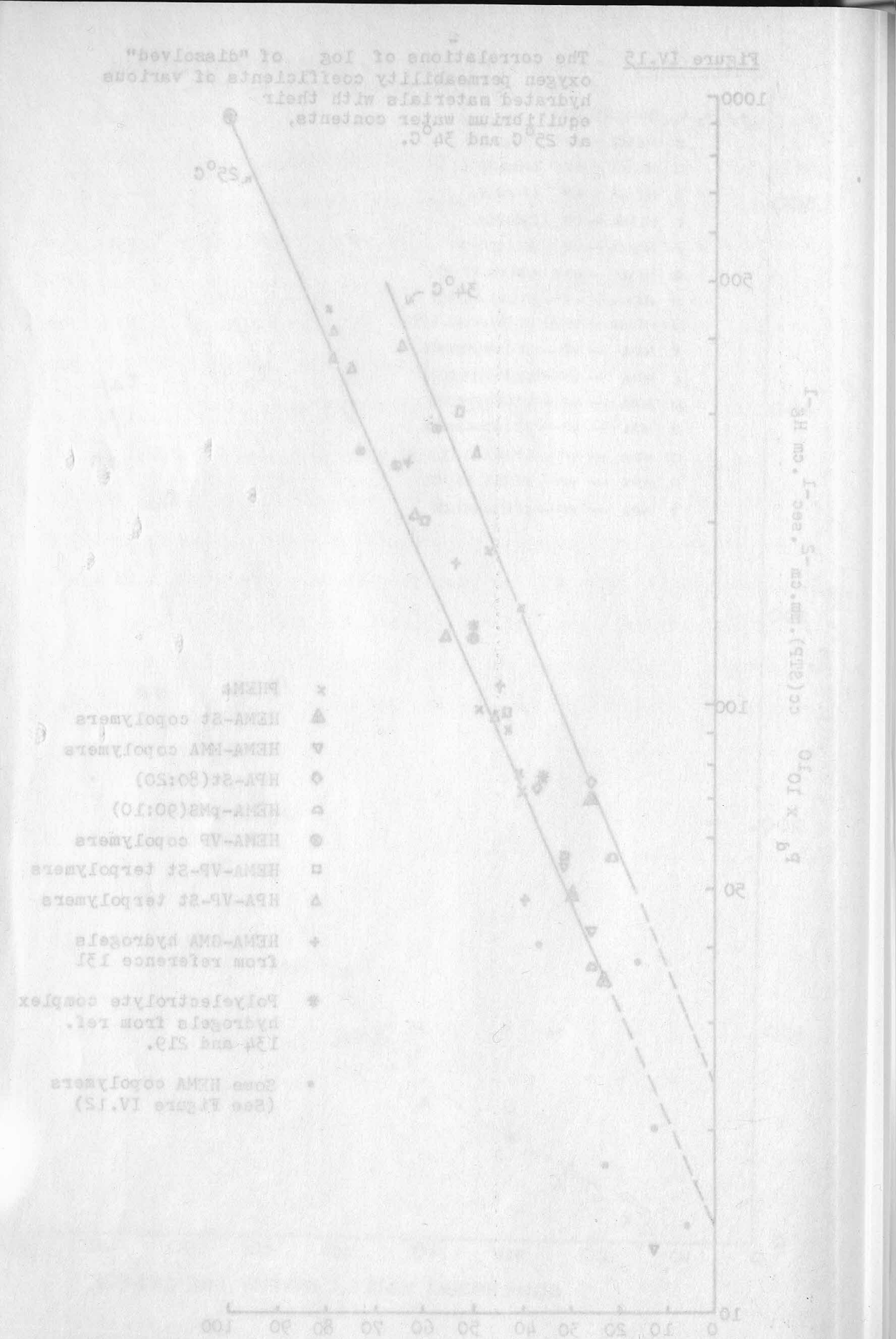


Figure IV.15 The correlations of log of "dissolved" oxygen permeability coefficients of various hydrated materials with their equilibrium water contents, at 25°C and 34°C.





IV.3.C. Theoretical Prediction of Required Contact Lens Thickness from "Dissolved" Oxygen Permeability Coefficient Measurements

It has been found[64] that in an open eye, a minimum oxygen tension of 11 to 19 mm.Hg is required at the anterior surface of the corneal epithelium for the cornea to show no sign of cornea swelling. This is expected to be also the critical oxygen tension under a contact lens during contact lens wear either in an open eye or a closed eye. It has been reported[61] that the cornea consumes oxygen at such a rate that in the case of a tight contact lens, the oxygen flux (F) across the epithelial surface is given by $F = \alpha p_2^{\beta}$, where α and β are two empirical constants ($\alpha = 0.24 \times 10^{-6}$ cc $O_2 . cm^{-2} . sec^{-1} . (mm\ Hg)^{-\frac{1}{2}}$ and $\beta = 0.5$) and p_2 the oxygen tension at the epithelial surface. For a critical oxygen tension of 15 mm Hg (average value) at the epithelial surface, the oxygen flux through a contact lens should be 0.93×10^{-6} cc (S.T.P.). $cm^{-2} . sec^{-1}$ which is about 3.5 $\mu l . cm^{-2} . hr^{-1}$.

Knowing the minimum oxygen flux through a contact lens to give the minimum oxygen tension of 15 mm Hg under the lens, it is possible to determine the required contact lens thickness (L) for a given "dissolved" oxygen permeability coefficient (P_d) for both open-eye and closed-eye situations, using Equation IV.13. The P_d and L correlations for a contact lens in an open-eye condition (i.e. Δp = 14.0 cm Hg) and in a closed-eye

$$F = \frac{P_d}{L} \times \Delta p \qquad \text{(IV.13)}$$

condition (i.e. Δp = 4.0 cm Hg) are shown in Figure IV.16. It shows the required minimum "dissolved" oxygen permeability coefficient of a contact lens material for a contact lens of any thickness to be made, or the maximum thickness of a contact lens if its "dissolved" oxygen permeability coefficient is known, in either open-eye or closed-eye condition. For a continuous-wear contact lens which is to be worn in

both open-eye and closed-eye conditions, the correlation between P_d and L for the closed-eye condition should be used. So for a continuous-wear contact lens of thickness 0.1 mm., the contact lens material should have a "dissolved" oxygen permeability coefficient at least 230 x 10^{-10} cc(S.T.P.).mm.cm^{-2}.sec^{-1}.cm Hg^{-1}, in order that the oxygen tension under the lens is maintained above the critical value of 15 mm Hg.

Hill and Fatt[54] have measured the oxygen flux into the cornea of an open eye in the atmospheric conditions to be about 7 µl.cm^{-2}.hr^{-1}, which is twice of the minimum oxygen flux calculated here. This means that if the value of 7 µl.cm^{-2}.hr^{-1} is used for the minimum oxygen flux, then the required "dissolved" oxygen permeability coefficient should be twice, or the contact lens thickness half, of that determined previously in the case of 3.5 µl.cm^{-2}.hr^{-1}, as shown in Figure IV.16. However an oxygen flux of 3.5 µl.cm^{-2}.hr^{-1} seems to be sufficient to maintain corneal transparency.

Alternatively for a known "dissolved" oxygen permeability coefficient (P_d) and a known contact lens thickness (L), the oxygen flux (F) through the contact lens can be predicted for any pressure difference across it, using Equation IV.13. The correlations between the oxygen flux (F) and the inverse of contact lens thickness, i.e. (1/L), for various hydrogel materials, poly(4-methyl pent-1-ene) and silicone rubber in both closed-eye and open-eye conditions are shown in Figures IV.17a and b. As shown in Figure IV.17a, PHEMA hydrogel having an equilibrium water content of 40% and a "dissolved" oxygen permeability coefficient of 145 x 10^{-10} cc (S.T.P.)mm.cm^{-2}.sec^{-1}. cm Hg^{-1} at 34°C would give an oxygen flux just sufficient for the corneal requirement in the open-eye condition but insufficient in the closed-eye condition, if the PHEMA contact lenses were 0.2 mm. thick. This illustrates the fact

Figure IV.16 Theoretical prediction of required contact lens thickness from the "dissolved" oxygen permeability coefficient for an oxygen flux of 3.5 or 7.0 $\mu l.cm^{-2}.hr^{-1}$ under both open-eye and closed-eye conditions, at 34°C.

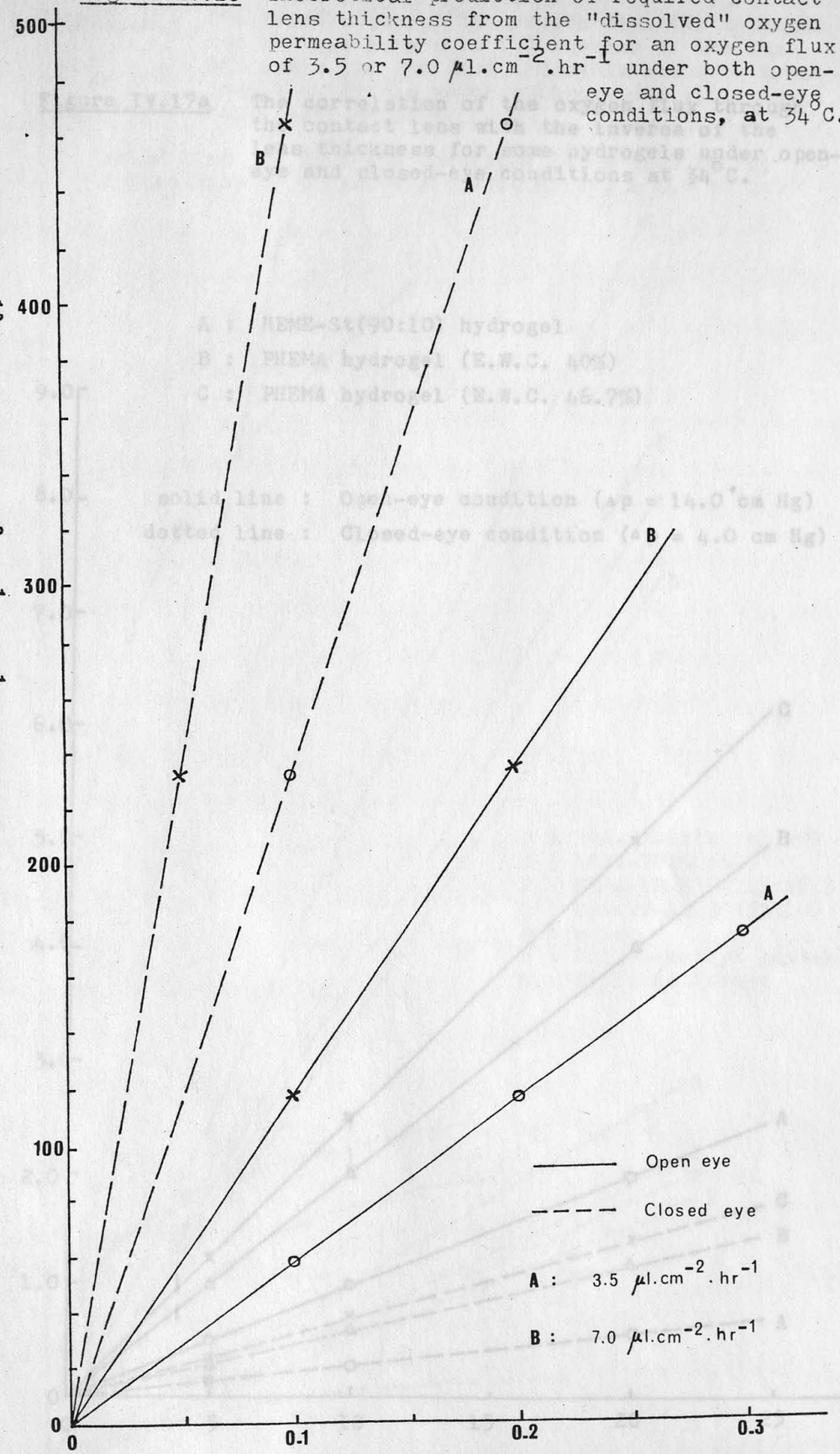


Figure IV.17a The correlation of the oxygen flux through the contact lens with the inverse of the lens thickness for some hydrogels under open-eye and closed-eye conditions at 34°C.

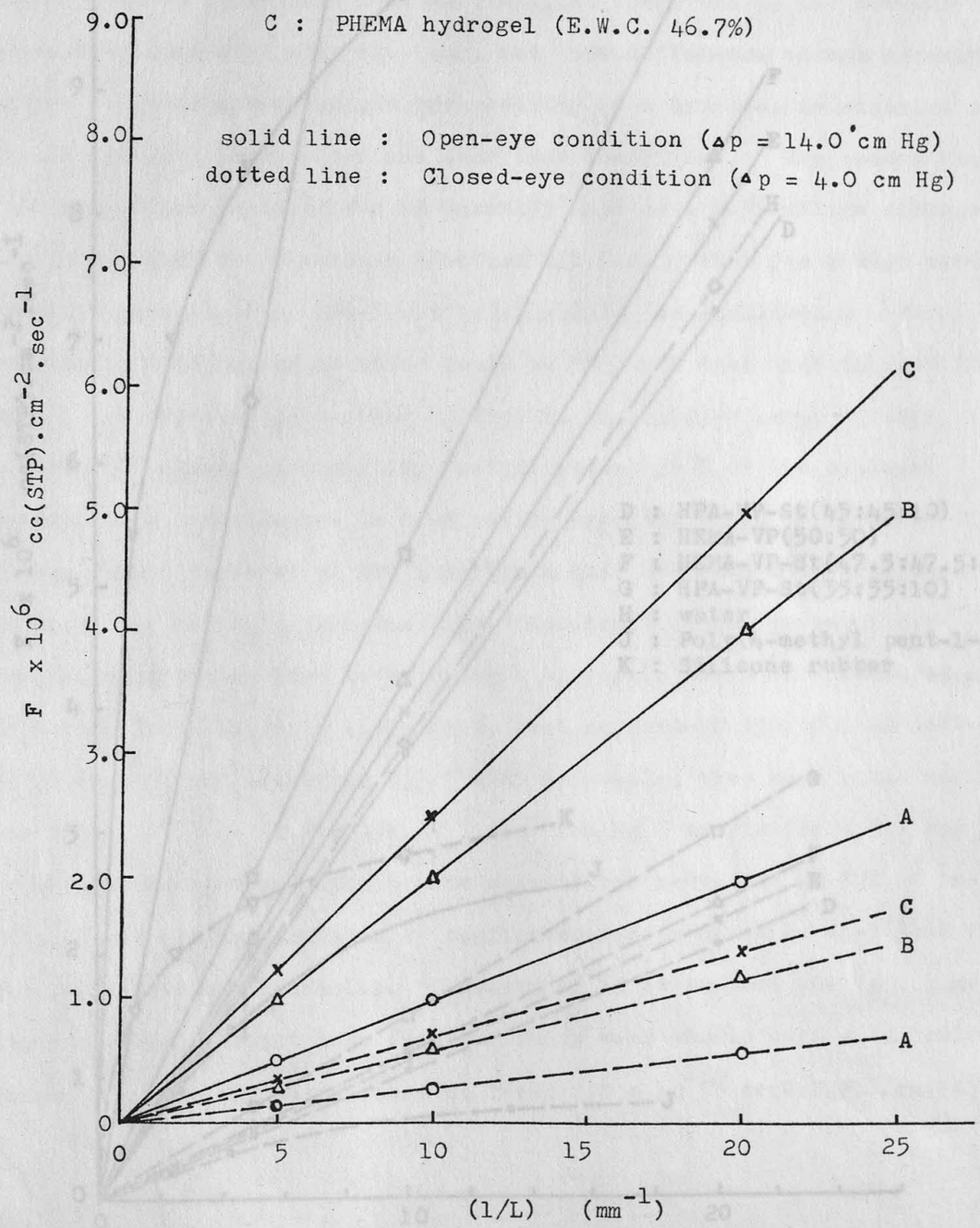
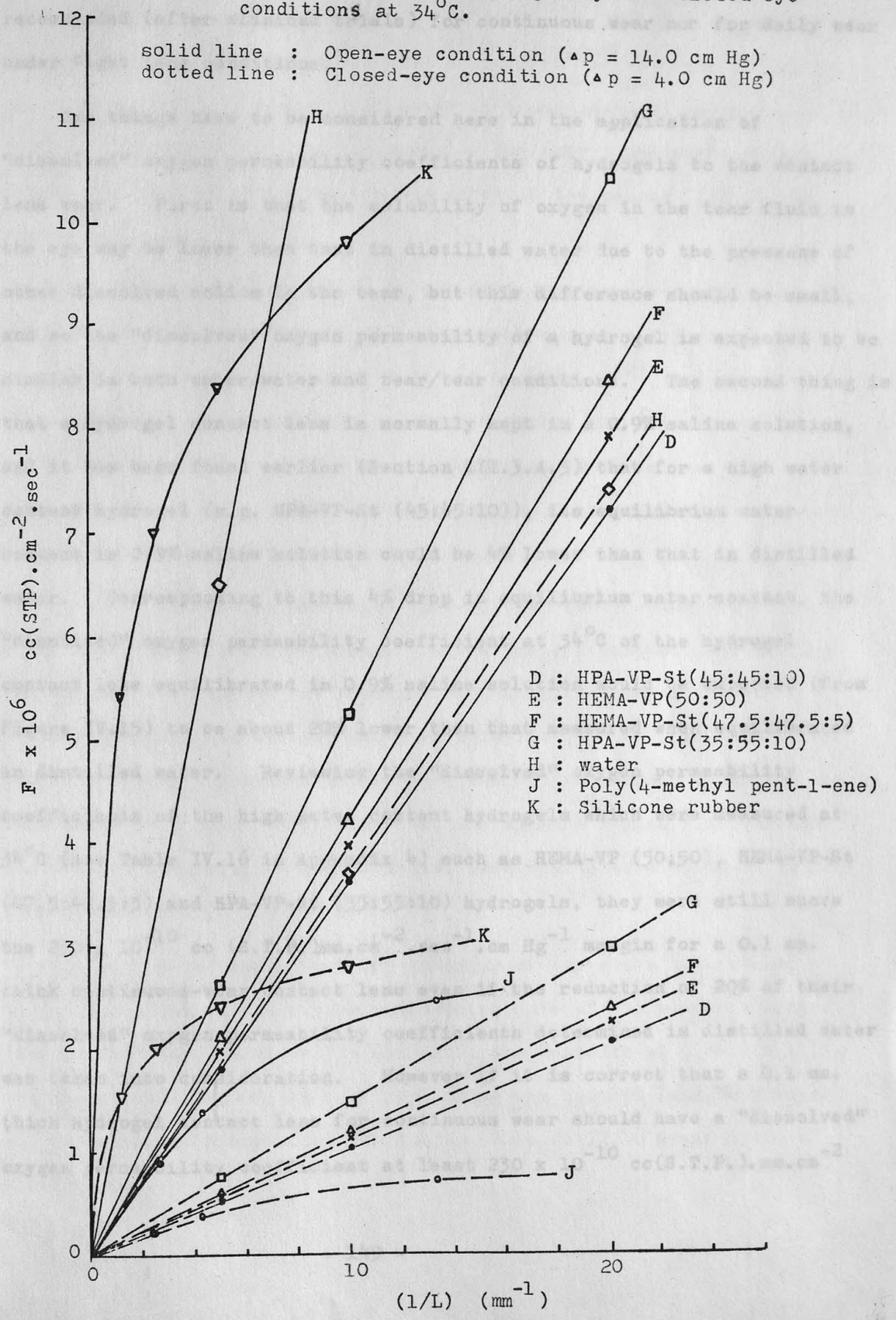
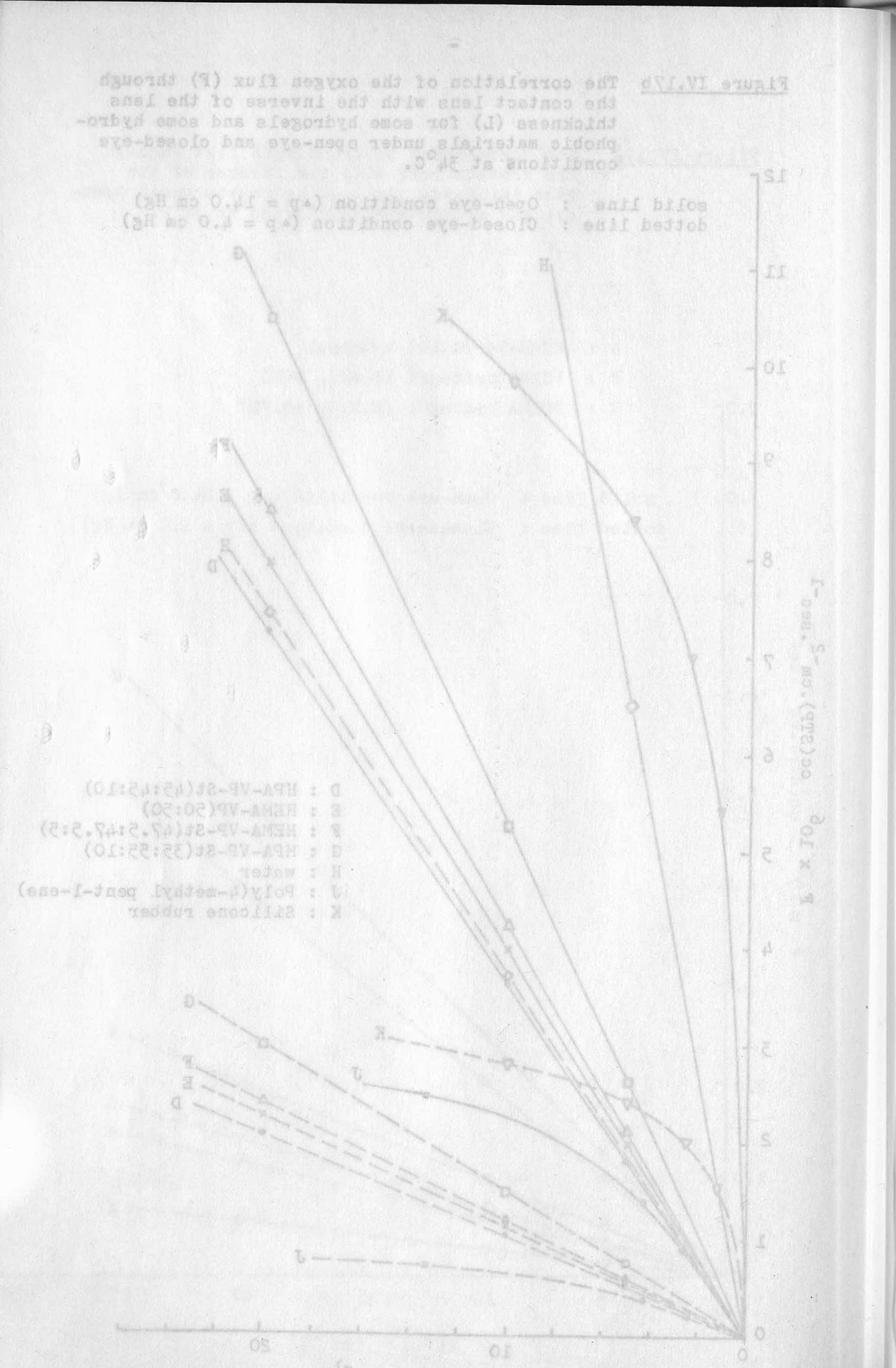


Figure IV.17b The correlation of the oxygen flux (F) through the contact lens with the inverse of the lens thickness (L) for some hydrogels and some hydrophobic materials under open-eye and closed-eye conditions at 34°C.





that commercial PHEMA hydrogel contact lenses of 0.3 mm. thick are not recommended (after clinical trials) for continuous wear nor for daily wear under tight lens conditions.

Two things have to be considered here in the application of "dissolved" oxygen permeability coefficients of hydrogels to the contact lens wear. First is that the solubility of oxygen in the tear fluid in the eye may be lower than that in distilled water due to the presence of other dissolved solids in the tear, but this difference should be small, and so the "dissolved" oxygen permeability of a hydrogel is expected to be similar in both water/water and tear/tear conditions. The second thing is that a hydrogel contact lens is normally kept in a 0.9% saline solution, and it has been found earlier (Section III.3.A.5) that for a high water content hydrogel (e.g. HPA-VP-St (45:45:10)), its equilibrium water content in 0.9% saline solution could be 4% lower than that in distilled water. Corresponding to this 4% drop in equilibrium water content, the "dissolved" oxygen permeability coefficient at 34°C of the hydrogel contact lens equilibrated in 0.9% saline solution would be expected (from Figure IV.15) to be about 20% lower than that measured when equilibrated in distilled water. Reviewing the "dissolved" oxygen permeability coefficients of the high water content hydrogels which were measured at 34°C (see Table IV.16 in Appendix 4) such as HEMA-VP (50;50), HEMA-VP-St (47.5:47.5:5) and HPA-VP-St (35:55:10) hydrogels, they were still above the 230×10^{-10} cc (S.T.P.)mm.cm^{-2}.sec^{-1}.cm Hg^{-1} margin for a 0.1 mm. thick continuous-wear contact lens even if the reduction of 20% of their "dissolved" oxygen permeability coefficients determined in distilled water was taken into consideration. However if it is correct that a 0.1 mm. thick hydrogel contact lens for continuous wear should have a "dissolved" oxygen permeability coefficient at least 230×10^{-10} cc(S.T.P.).mm.cm^{-2}

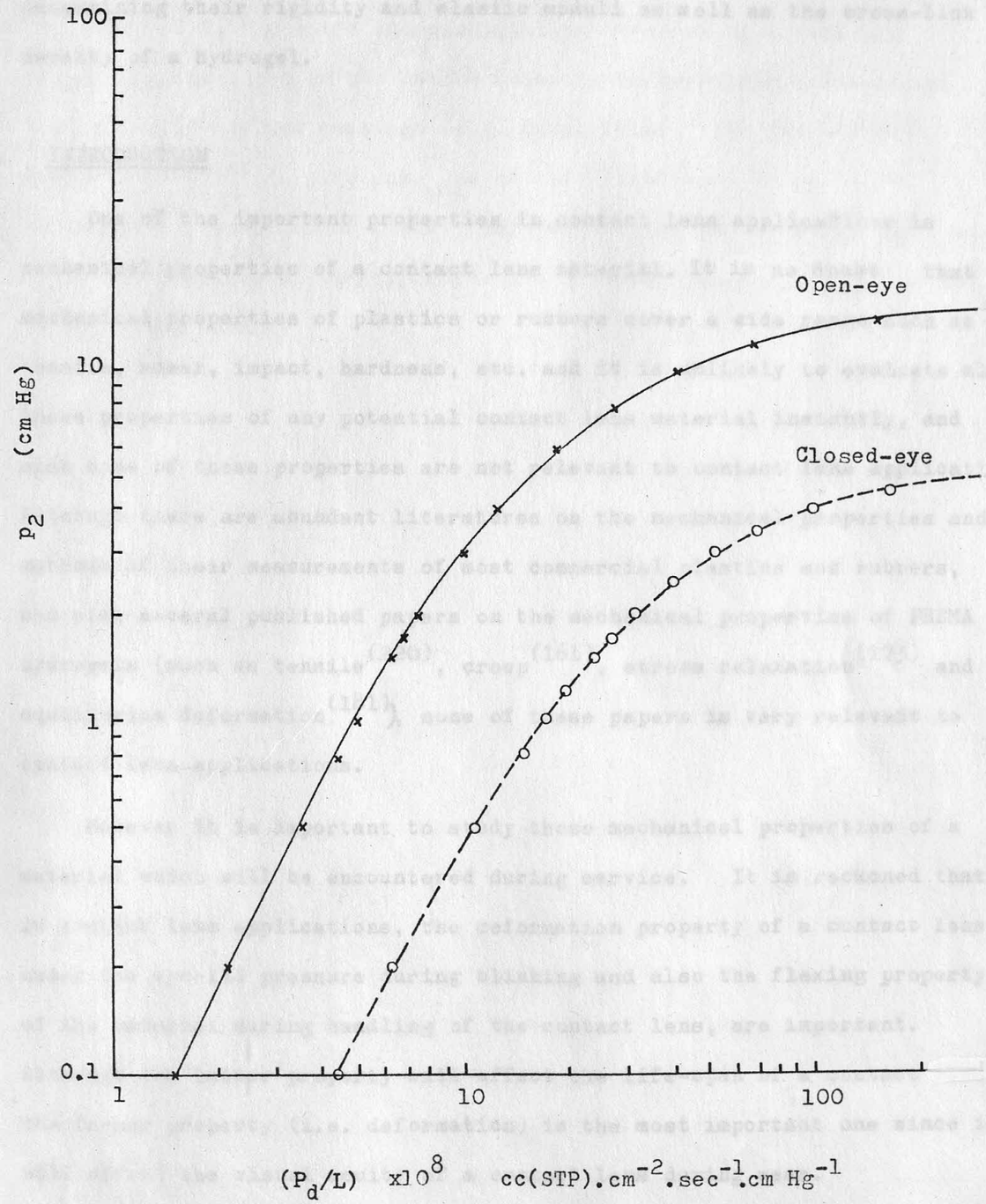
sec^{-1}. cm Hg^{-1}, then the lens should have an equilibrium water content at least 52% in 0.9% saline solution at 34°C (Figure IV.15). Recently Holly and Refojo[214] estimated that a hydrogel contact lens of 0.1 mm. thick should contain roughly 86% water in order to be so permeable as to satisfy the normal oxygen requirement of the cornea under closed-eye conditions, but their estimation was based on the oxygen flux of the cornea of 7 $\mu l.cm^{-2}.hr^{-1}$.

An alternative approach is to correlate the oxygen tension under a contact lens and the ratio (P_d/L) under both open-eye and closed-eye conditions, as has been done by Fatt and St.Helen[136], and shown in Figure IV.18. So by determining experimentally the "dissolved" oxygen permeability coefficient (P_d) of a contact lens material and the thickness (L) of the contact lens, the oxygen tension under the contact lens during wear can be predicted.

However it must be mentioned that all the above considerations are true under tight lens conditions. Since present contact lens fitting techniques give rise to the presence of a thin tear layer between a contact lens and the cornea, the pumping action of the tear pool under the lens during blinking will increase the oxygen tension under the lens. This means that under this situation, the minimum oxygen flux through the contact lens will be less critical than that discussed under tight lens conditions.

It must be stressed that the validity of the above theoretical approach depends on the oxygen consumption rate of the wearer's cornea (varies from one individual to another), the validity of Equation IV.13 and the circulation of the tear pool under the contact lens. However all these correlations shown in Figures IV.16 to IV.18 await clinical confirmation.

Figure IV.18 The prediction of the oxygen tension (p_2) under a tight contact lens from its (P_d/L) value. (Reference 136)



Chapter V. DEFORMATION PROPERTIES OF HYDROGELS

This chapter covers deformation properties of some hydrogels and some hydrophobic polymers under static loads using the I.C.I. Micro-indentation Apparatus, together with its potential applications in determining their rigidity and elastic moduli as well as the cross-link density of a hydrogel.

V.1. INTRODUCTION

One of the important properties in contact lens applications is mechanical properties of a contact lens material. It is no doubt that mechanical properties of plastics or rubbers cover a wide range such as tensile, shear, impact, hardness, etc. and it is unlikely to evaluate all these properties of any potential contact lens material instantly, and also some of these properties are not relevant to contact lens applications. Although there are abundant literatures on the mechanical properties and methods of their measurements of most commercial plastics and rubbers, and also several published papers on the mechanical properties of PHEMA hydrogels (such as tensile[220], creep[161], stress relaxation[173] and equilibrium deformation[181]), none of these papers is very relevant to contact lens applications.

However it is important to study those mechanical properties of a material which will be encountered during service. It is reckoned that in contact lens applications, the deformation property of a contact lens under the eye-lid pressure during blinking and also the flexing property of the material during handling of the contact lens, are important. Although the latter property will affect the life-span of a contact lens, the former property (i.e. deformation) is the most important one since it will affect the visual acuity of a contact lens during wear.

In search of a commercial instrument to study deformation properties of contact lens materials, particularly hydrogels under compressive loads, the I.C.I. Pneumatic Microindentation Hardness Apparatus was found to be most suitable since it requires only a small test sample (similar to the size of a contact lens).

The I.C.I. Pneumatic Microindentation Hardness Apparatus was originally developed in the Paints Industry to correlate deformation properties of surface coatings (e.g. paint films) with their ageing properties[221,222]. To make use of this instrument to study the deformation properties of contact lens materials, particularly hydrogels, some modifications of the apparatus are necessary in this work, and will be described later.

V.2. EXPERIMENTAL

V.2.A. Description of the I.C.I. Pneumatic Microindentation Hardness Apparatus

The I.C.I. Pneumatic Microindentation Hardness Apparatus (made under license by the Research Equipment Ltd.) consists of two units, i.e. the basic instrument and the pneumatic measuring and recording unit.

V.2.A.1. Basic Instrument

The basic instrument is shown in Figure V.1. It consists of a triangular steel plate A, which is supported from the base plate X by means of three adjustable legs C, D and E. The legs C and D form the coarse and fine adjustments respectively for raising and lowering the top plate A during zeroing and calibration. The wheel D is divided into 1-micron divisions so that one revolution of this wheel gives a movement of 20 microns at the specimen. A beam M_1 is mounted on a crossed fluxure bearing G (made of brass) and joined to an adjustable weight H, and on the other end of the beam are an oil dash pot J, an indentor needle I and a flapper F_1 of a flapper/nozzle system. A specimen S is held down by a spring N on a metal stage, which sits on a T.E.C.H. unit T (thermoelectric Cooling/Heating Unit) consisting of a Frigistor and a bigger metal stage through which a stream of tap water runs. The Frigistor is connected to a low voltage D.C. electric supply and monitor (A.C. mains operated) with a controllable output of 0-15 amp., and can provide any temperature between -20°C and 90°C in a few minutes. The temperature of the test specimen is measured with a thermocouple connected from a sample stage to a pyrometer (Cambridge).

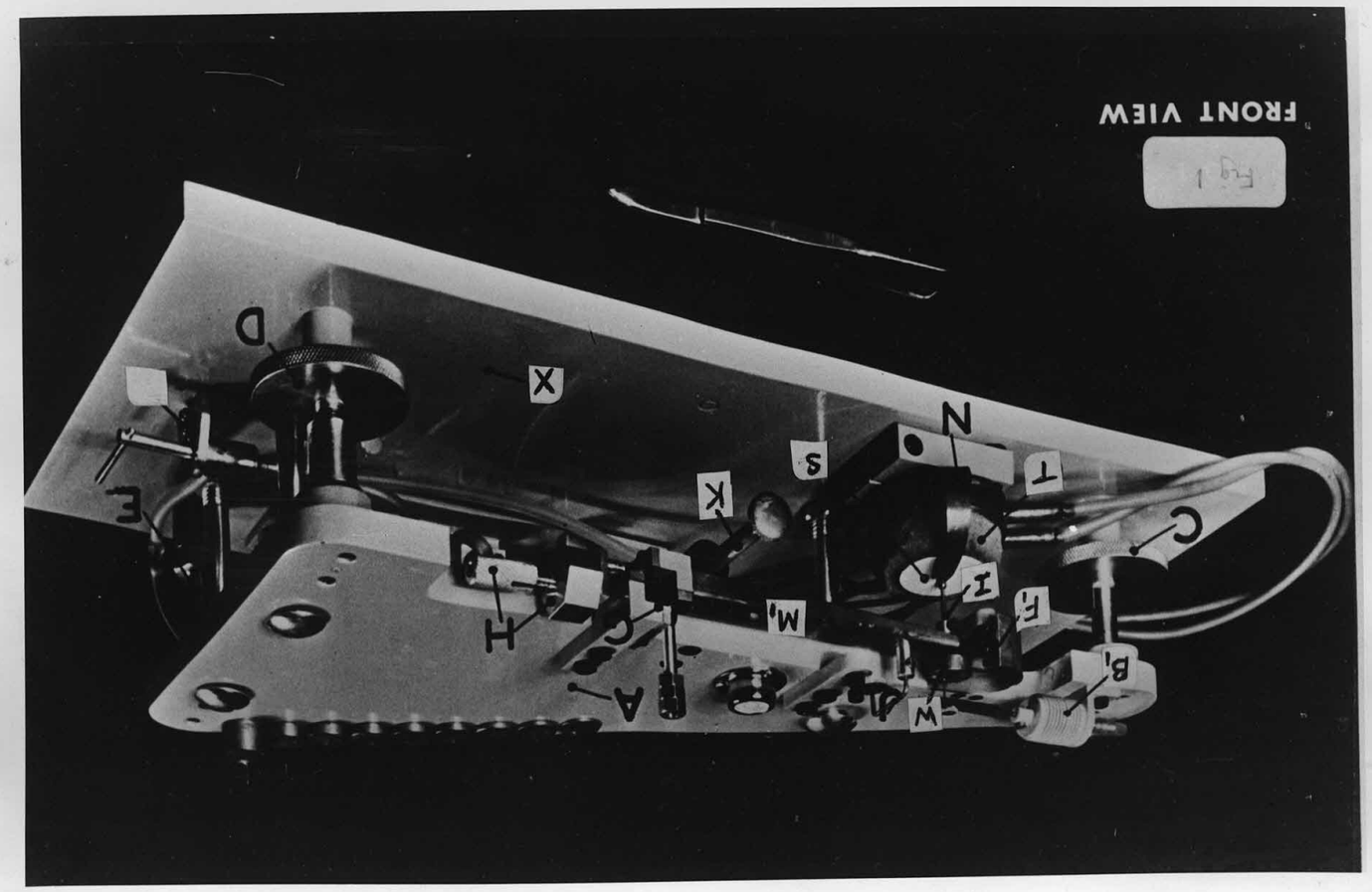


Figure V.1 The basic instrument of the I.C.I. Micro-indentation Hardness Apparatus.

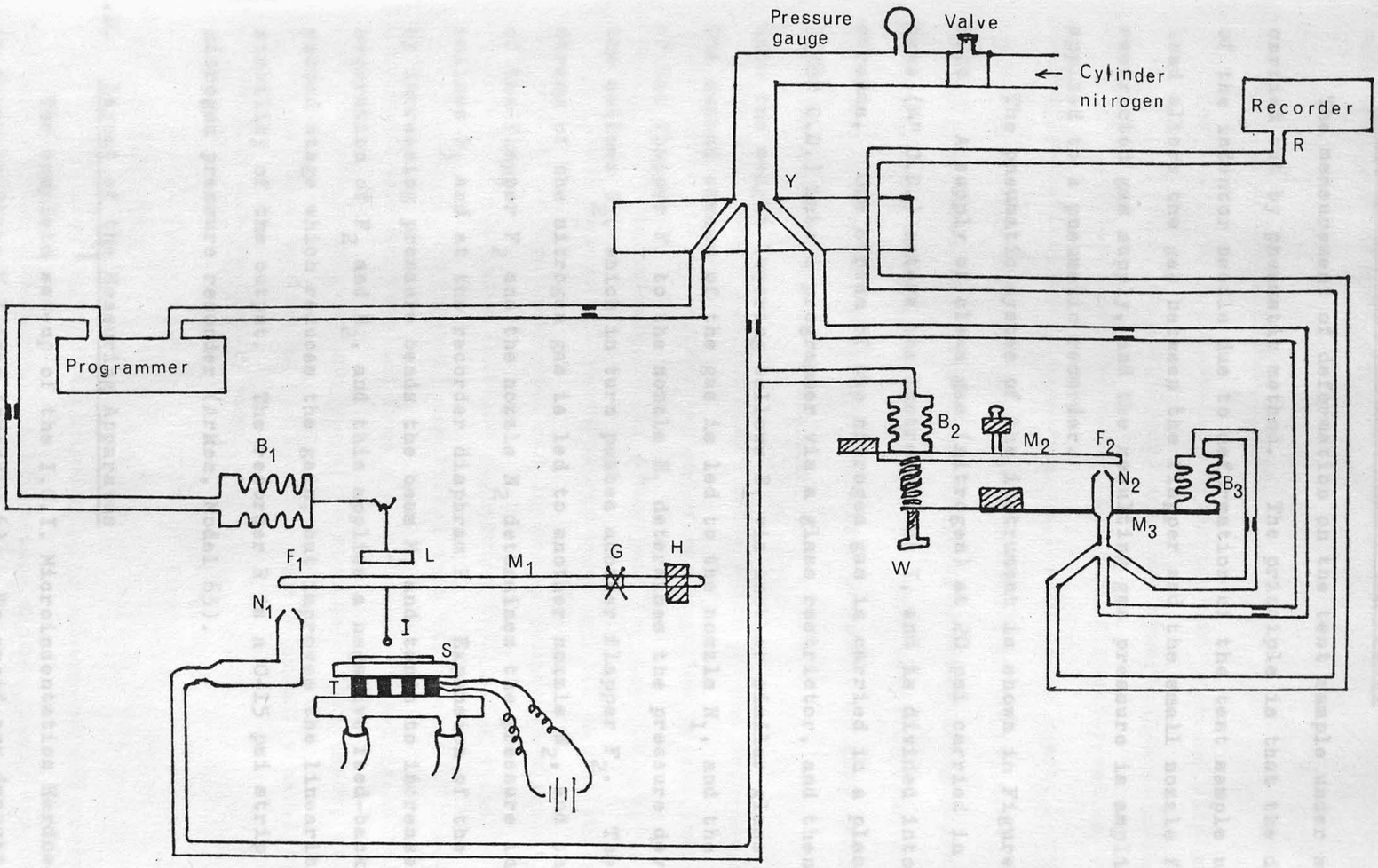


FIGURE V.2b **SCHEMATIC DIAGRAM OF THE PNEUMATIC SYSTEM**

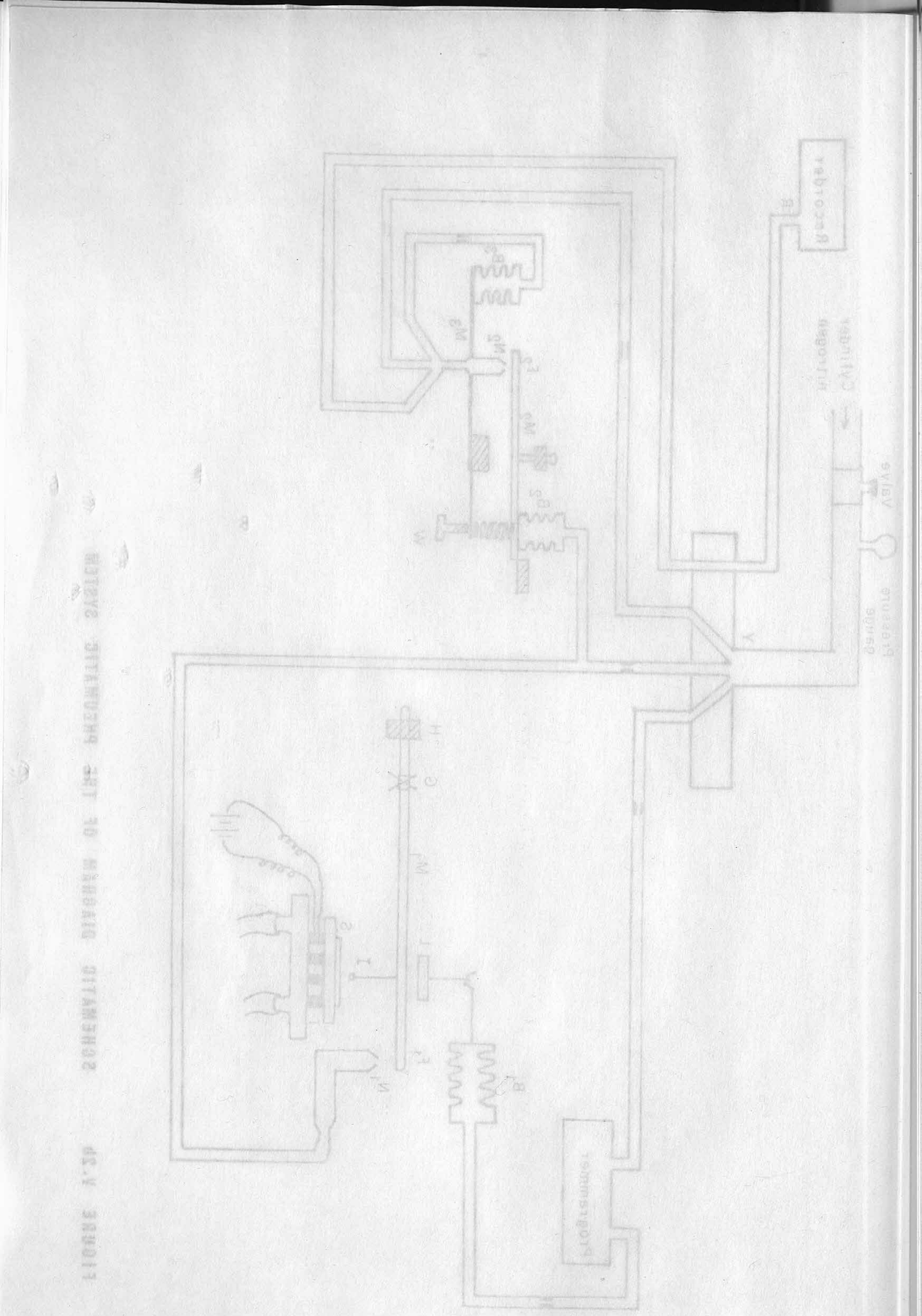


FIGURE V.2b SCHEMATIC DIAGRAM OF THE PNEUMATIC SYSTEM

V.2.A.2. The Pneumatic Measuring and Recording System

The measurement of deformation on the test sample under a load is carried out by pneumatic method. The principle is that the displacement of the indentor needle due to deformation of the test sample under the load alters the gap between the flapper and the small nozzle fed with a restricted gas supply, and the resulting gas pressure is amplified and applied to a pneumatic recorder.

The pneumatic system of the instrument is shown in Figures V.2a and V.2b. A supply of clean gas (nitrogen) at 20 psi carried in a plastic tube (¼" O.D.) enters the instrument at Y, and is divided into three streams, One stream of the nitrogen gas is carried in a plastic tube (1/8" O.D.) into a programmer via a glass restrictor, and then out again into the weight lowering bellows B_1 via another similar glass restrictor. The second stream of the gas is led to the nozzle N_1, and the proximity of the flapper F_1 to the nozzle N_1 determines the pressure developed in the bellows B_2, which in turn pushes another flapper F_2. The third stream of the nitrogen gas is led to another nozzle N_2, and the proximity of the Flapper F_2 and the nozzle N_2 determines the pressure in the bellows B_3 and at the recorder diaphram R. Expansion of the bellows B_3 by increasing pressure bends the beam M_3 and tends to increase the separation of F_2 and N_2, and this applies a negative feed-back to the second stage which reduces the gain, but improves the linearity and stability of the output. The recorder R is a 0-15 psi strip chart nitrogen pressure recorder (Arkon, Model 63).

V.2.B. Layout of the Measuring Apparatus

The complete set-up of the I.C.I. Microindentation Hardness Apparatus is shown in Plate V.1 (see Appendix 6). To avoid any draught during measurement, the basic instrument is housed in a glove box (made of perspex) which has two "glove" holes in front of the box, but no gloves

were used in this work.

V.2.C. Calibration and Testing Procedure

The I.C.I. Microindentation Apparatus was provided with various loads (1, 2, 4, 8, 14 and 32 gms) and four spherical-end indentors (1/16 and 1/64 inch diameter, two each), and was originally set up to give a full-scale deflection of 6 microns on the recorder chart. For the purposes of this work, weights smaller than 1 gm (0.14, 0.24, 0.34, 0.48 and 0.75 gm) were made from wires (e.g. by straightening paper clips), and the instrument was also calibrated with other sensitivities by adjusting the screw W near the bellows B_2 of the instrument to give larger full-scale deflections up to 30 microns.

The programmer and the chart drive motor of the recorder were connected to the mains, and tap water was supplied to the instrument at about 1 litre per minute. The power pack for the T.E.C.H. unit was not used in this work as all measurements were carried out at room temperature. The beam M_1 of the instrument was locked by gently turning the knob K (Figure V.1) clockwise (without tightening it), and a suitable indentor (spherical-end or flat-end) was fitted into the socket of the beam. A nitrogen gas was supplied from a cylinder at 20 psi (read from the provided precision pressure gauge) to the whole instrument, and the beam M_1 was released by turning the locking knob K slightly until the hole on the knob was completely seen.

V.2.C.1. Calibration

The calibration of the instrument for subsequent deformation studies was carried out as follows. A microscope slide (glass) was placed on the specimen stage and secured with the clip. A sensitivity of the instrument was preset by adjusting the screw W, and the counterbalance

weight H was adjusted until a full-scale deflection was obtained on the recorder chart. A load 16 gm (or 0.34 gm) was placed on the hook of the weight lowering bellows B_1 and then lowered on the beam M_1 by switching the programmer to "TEST LOAD". The fine adjusting wheel D was set at zero by rotating in an anti-clockwise direction. The top plate A of the instrument was lowered by the coarse adjusting wheel C (clockwise) until the indentor needle touched the glass slide and the recorder indicated zero on the chart. The fine adjusting wheel was then rotated in an anti-clockwise direction at every interval of one micron, and the corresponding micron reading on the recorder chart was taken. A calibration curve was then constructed for that sensitivity by plotting the recorder micron readings vs. the micron readings on the fine adjusting wheel.

After the calibration, the fine adjusting wheel was returned to zero, and the "TEST LOAD" was switched off. The indentor needle was raised to leave the glass specimen by the coarse adjusting wheel, and the glass specimen was then removed. If the recorder pen did not show the full-scale deflection, the counter-balance weight H was again adjusted. The instrument was then ready for deformation measurements of contact lens materials.

V.2.C.2. Deformation Measurements

Depending on the purpose of deformation measurements, different shapes and sizes of the indentor needle and different weights were used. In the case of deformation measurements under the eye-lid pressure, a flat-end indentor (0.126 cm. diameter at the flat end) and a weight of 0.34 gm. were used. In the case of deformation measurements to determine rigidity and elastic moduli of the materials, a spherical-end indentor

(1/16 inch, i.e. 0.160 cm.diameter) and various weights (from 0.14 gm. to 16 gms) were used. In the case of deformation measurements to determine the cross-link density of a hydrogel, the flat-end indentor and various weights were used. The flat-end indentor was obtained by breaking the spherical end of an indentor so that the broken end was fitted into the socket of the beam leaving the other end (flat) of the indentor free for contact with test samples during deformation measurements.

A hydrogel test sample of known thickness (about 1 cm. diameter disc) was placed on a microscope slide (about 1" x 1" square), ensuring a good contact without any bubbles trapped and the surface of the hydrogel sample was covered with a thin layer of paraffin oil to prevent evaporation of water during the deformation measurement. The microscope slide with the sample was then placed on the specimen stage and secured with the clip. The beam M_1 was lowered by means of the coarse and fine adjusting wheel until the indentor needle touched the test sample, and then only the fine adjusting wheel was adjusted until the recorder chart indicated zero. A suitable weight was put on the hook of the weight lowering bellows, and the programmer was switched to "START". The indentation and recovery curves were recorded automatically, and the time scale for each curve could be adjusted by a screw behind the programmer. (For all the measurements in this work, the time scale was set at 1 minute for each of the indentation and recovery.) Immediately after the completion of the recovery curve, the programmer was switched to "RESET" and then "START" again so that another indentation and recovery curves were recorded on the same spot of the test specimen. This again was repeated once or twice so that an average deformation on one spot of the test sample was obtained. To measure the deformation of another point of the test specimen under the same weight, the beam was lifted by the coarse adjusting wheel so that the indentor needle was clear off the specimen, and then the specimen was moved to a new spot and the deformation measurements were repeated as described above. Thus the

deformation and recovery characteristics of three or four points on the test specimen were obtained, and an average depth of indentation in microns on the test specimen under one load was calculated. In the experiments involving various loads, the deformation measurements were repeated with each load on the same sample (loads in the ascending order).

For comparisons, deformation measurements were also carried out on some hydrophobic materials such as poly(methyl methacrylate) (conventional contact lens material), polystyrene, poly(4-methyl pent-1-ene), ethylene propylene terpolymer, and dehydrated PHEMA and HEMA-St (90:10). Except in the case of ethylene propylene terpolymer which could adhere to a microscope slide quite well, all the other hydrophobic materials were glued to microscope slides with "Araldite" which was allowed to set at 40°C overnight.

In the determination of cross-link density of PHEMA hydrogel, a dehydrated PHEMA was equilibrated with dimethyl sulphoxide (volume swell, V_G/V_P = 585%), and the deformation measurements were carried out on this dimethyl sulphoxide-swollen sample (volume fraction of the polymer in the gel, $v_r = V_P/V_G$ = 0.171).

V.3. RESULTS AND DISCUSSIONS

V.3.A. Deformation Properties under the Eye-lid Pressure

It has been reported[45] that the eye-lid pressure during blinking is 2.6×10^4 dynes.cm^{-2}. Although the actual eye-lid forces on the contact lens may be complex, compressive forces seem to be the major contribution. So for simplicity, comparative deformations of various hydrogels were studied under a compressive force equivalent to the eye-lid pressure. Thus the 1-minute deformation and recovery characteristics of various hydrogels, and some hydrophobic materials for comparison under the eye-lid pressure (0.34 g. load and a flat-end indentor of 0.126 cm. diameter) are shown in Figures V.3 to V.16.

For each material, measurements were made at several points of the sample, and at each point, a set of 2 consecutive deformation/recovery curves were made. The curves were found to vary to some degrees between the sets for the same material, and only the best set, however, of 2 consecutive deformation/recovery curves for each material is shown (except for PHEMA hydrogel in Figure V.3 and for HPMA-VP (50:50) hydrogel in Figure V.11, where the curves of more sets are shown). As shown in the figures, there was always a shift of the base-line in the first deformation/recovery curve of each set of the measurements, possibly due to the settling of the sample, which was not firmly placed on the sample support (i.e. microscope slide). So only the second deformation/recovery curve of each set was used for comparisons among various material

The "kink" observed at the beginning of the deformation curve was due to the change in nitrogen pressure of the system when the loading was switched on. There were difficulties sometimes in obtaining a good curve for a material, e.g. (1) a sharp decrease was observed in the curve due

to the drying-up of the sample at the edge, as shown in Figure V.11(set 1) or (2) an increase in the recovery curve (instead of the normal decrease) was obtained, as shown in Figure V.11 (set 2), due to the rise of the base-line. The latter was normally observed in the cases of high-water-content hydrogels. Thus little information could be derived from the distorted part of the deformation/recovery curve, especially the recovery portion.

Despite the difficulties encountered in obtaining reasonably good and reproducible deformation/recovery curves for all the materials, several observations were made from the results obtained about the shapes of the deformation and recovery curves, as well as the 1-minute deformation.

PHEMA hydrogel (equilibrium water content ~39%) was found to give a fairly good elastic behaviour, as seen from both the deformation and recovery curves (Figure V.3). As the load was applied on the sample, the sample gave an instantaneous deformation (i.e. the straight-line portion of the deformation curve), followed by a small retarded deformation. As the load was removed, there was an instantaneous recovery, followed by a small retarded recovery. Replacing 10% HEMA by MMA had little effect on the shapes of the deformation and recovery curves of the hydrogel (i.e. HEMA-MMA (90:10), equilibrium water content ~31%), except a small decrease in the 1-minute deformation (Figure V.4). On the other hand, replacing 10% HEMA by styrene resulted in a smaller instantaneous deformation but more time-dependency of both the deformation and recovery characteristics in HEMA-St (90:10) hydrogel (equilibrium water content ~23%), i.e. HEMA-St (90:10) hydrogel showed more viscoelastic properties than PHEMA hydrogel, but a smaller 1-minute deformation (Figure V.5). The deformation and recovery characteristics

of HPA-St (80:20) hydrogel (equilibrium water content ~28%) were found to show more viscoelasticity (Figure V.6) than PHEMA hydrogel, but the depth of the 1-minute deformation was similar. Incorporation of 1% EGDMA (a cross-linking agent) into HPA-St (80:20) reduced the 1-minute deformation and also increased the elasticity of the material, as shown in the deformation/recovery curve (Figure V.7). For the HEMA-EA (50:50) hydrogel (equilibrium water content ~ 16%), the 1-minute deformation and the shapes of the deformation/recovery curves were similar to those of PHEMA hydrogel, as shown in Figure V.8.

For higher water content hydrogels such as HEMA-VP (30:70) and HEMA-VP (50:50) hydrogels (equilibrium water contents ~50% and 66% respectively), larger 1-minute deformations than in the case of PHEMA hydrogel were obtained, and retarded deformations were also observed, as shown in Figures V.9 and V.10 respectively. HPMA-VP (50:50) hydrogel (equilibrium water content ~69%) showed a retarded deformation under the eye-lid pressure (Figure V.11), and gave a larger 1-minute deformation than PHEMA hydrogel, But incorporation of 1.2% EGDMA reduced the depth of the 1-minute deformation and also the time-dependency of both the deformation and recovery curves (Figure V.12).

For comparison, EPT was found to show an elastic deformation and an elastic recovery, and a smaller deformation than PHEMA hydrogel (Figure V.13). For poly(4-methyl pent-1-ene), polystyrene and PMMA, the deformations under the eye-lid pressure were both very small, and could hardly be seen in Figures V.14, V.15 and V.16 respectively.

All the above results indicate that, under the same experimental conditions, a hydrogel having a higher equilibrium water content had a larger deformation under the eye-lid pressure (for a copolymer series

having various compositions), and a hydrogel containing more flexible groups had a larger deformation than a hydrogel containing more rigid groups (e.g. compare HEMA-EA (50:50) and HEMA-St (90:10) hydrogels, though the former had a lower equilibrium water content). Rigid plastics such as PMMA behaved like hard rubbers, i.e. small and elastic deformation, and complete recovery. Increasing the equilibrium water contents of the hydrogels increased the instantaneous deformation and instantaneous recovery, and for hydrogels of a copolymer series (e.g. HEMA-St copolymers), the retarded deformation (and also the retarded recovery) seemed to go through a maximum at some copolymer composition (compare the smaller retarded deformations for polystyrene and PHEMA hydrogel with that of HEMA-St (90:10) hydrogel). Those hydrogels having an equilibrium water content higher than 40% to 50% (e.g. HEMA-VP (50:50) hydrogel) seemed to have larger retarded deformations than PHEMA hydrogel unless they contained a 1% cross-linking agent. The increasing retarded deformations observed in the high-water-content hydrogels might be due to the inelastic property of water. Cross-linking reduced the equilibrium water content of a hydrogel and also increased the elasticity of the polymer network, resulting in a smaller instantaneous deformation and instantaneous recovery, as well as a much smaller retarded deformation and retarded recovery.

The difference between the starting point of the deformation curve and the final point of the recovery curve could possibly arise from (a) the permanent deformation of the material, and/or (b) the incompleteness of the retarded recovery within the experimental time period, and/or (c) the shift of the base-line. Although it was difficult to know the exact reason(s) of such a difference in each of the figures shown, it seemed that the recovery of some of the hydrogels after the removal of the load (especially high-water-content hydrogels) was not complete but left a small permanent deformation.

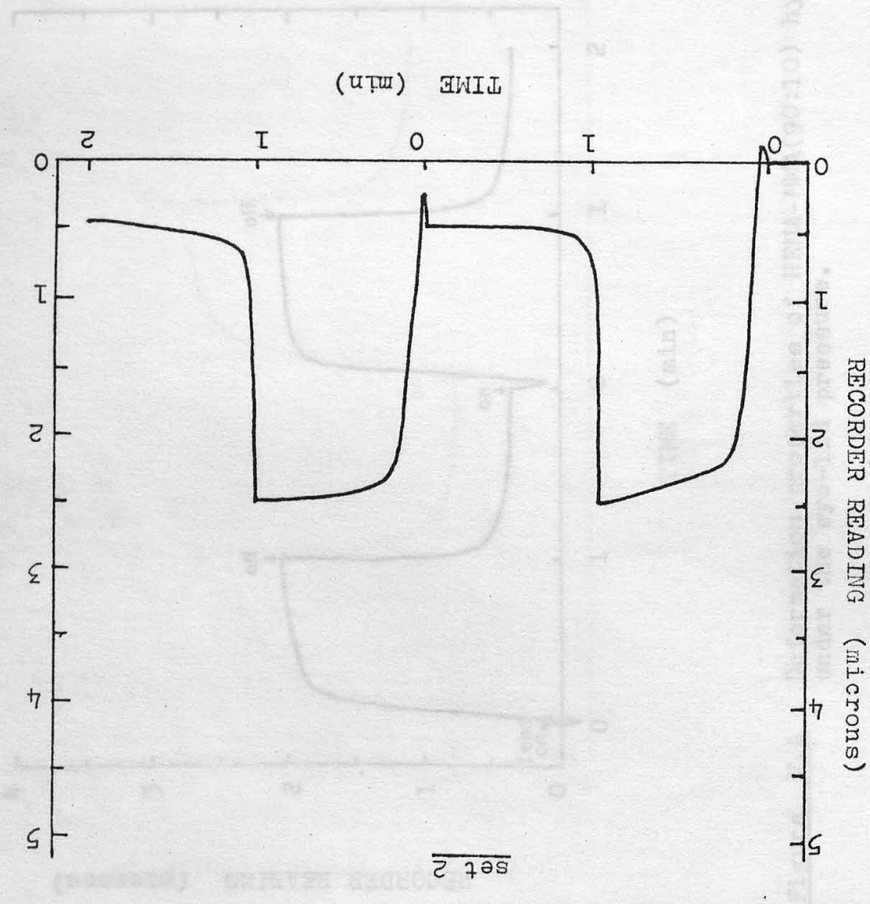


Figure V.3 Deformation properties of PHEMA hydrogel under the eye-lid pressure.

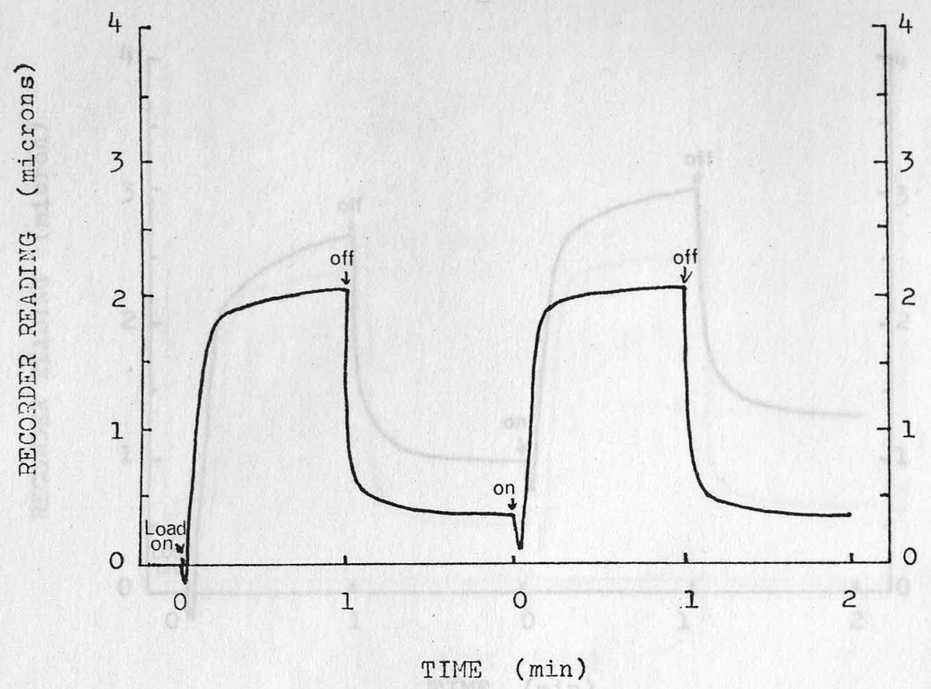


Figure V.4 Deformation properties of HEMA-MMA(90:10) hydrogel under the eye-lid pressure.

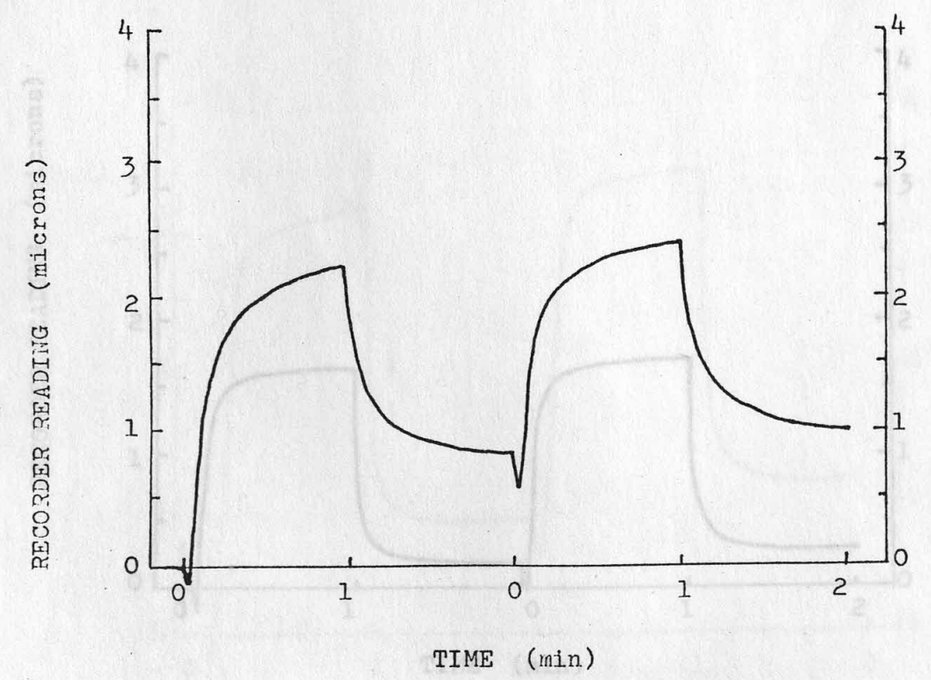


Figure V.5 Deformation properties of HEMA-St(90:10) hydrogel under the eye-lid pressure.

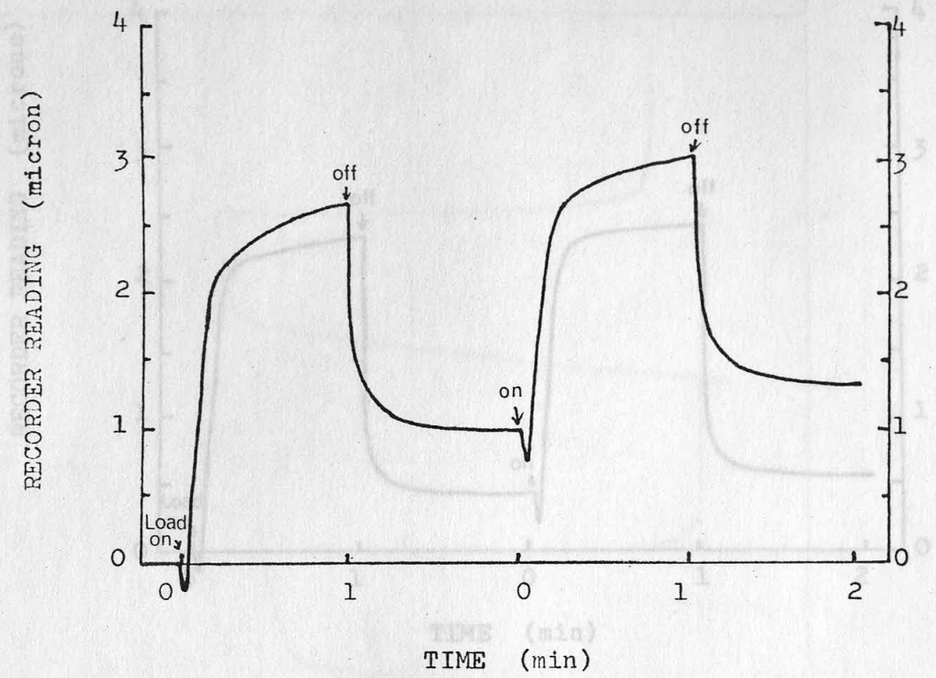


Figure V.6 Deformation properties of HPA-St(80:20) hydrogel under the eye-lid pressure.

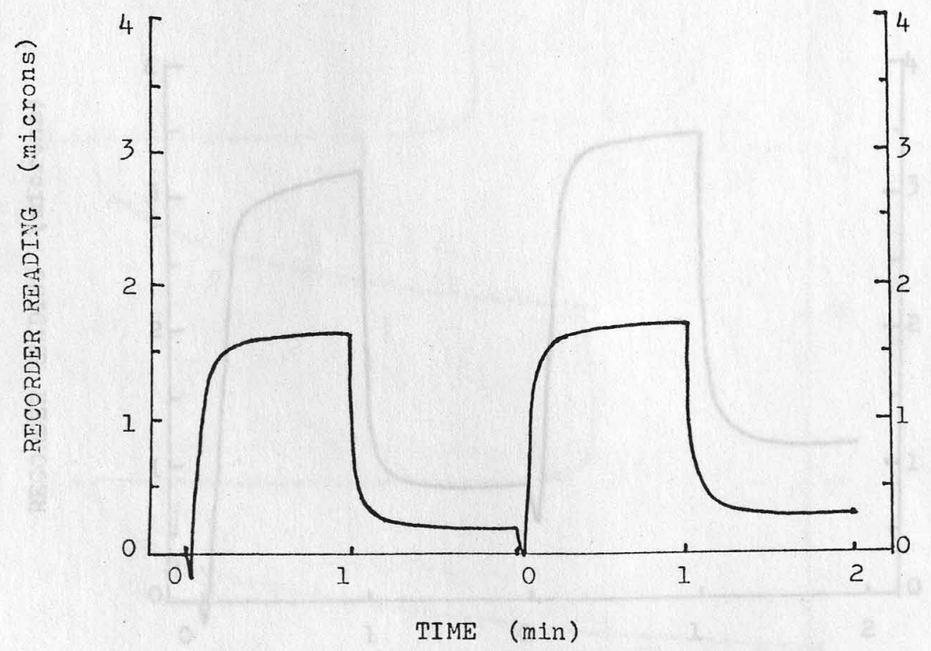


Figure V.7 Deformation properties of HPA-St(80:20) + 1% EGDMA hydrogel under the eye-lid pressure.

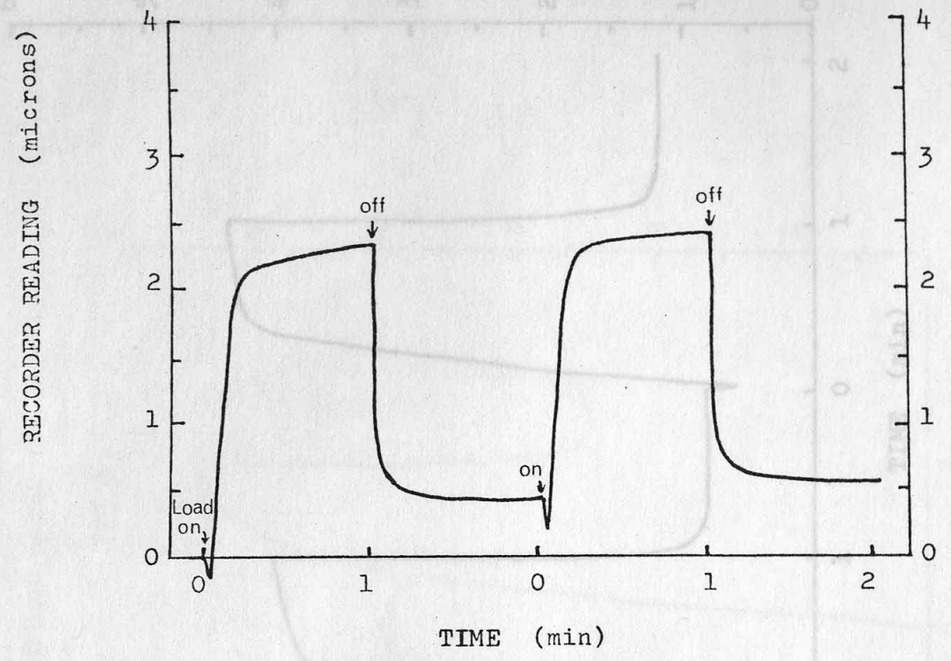


Figure V.8 Deformation properties of HEMA-EA(50:50) hydrogel under the eye-lid pressure.

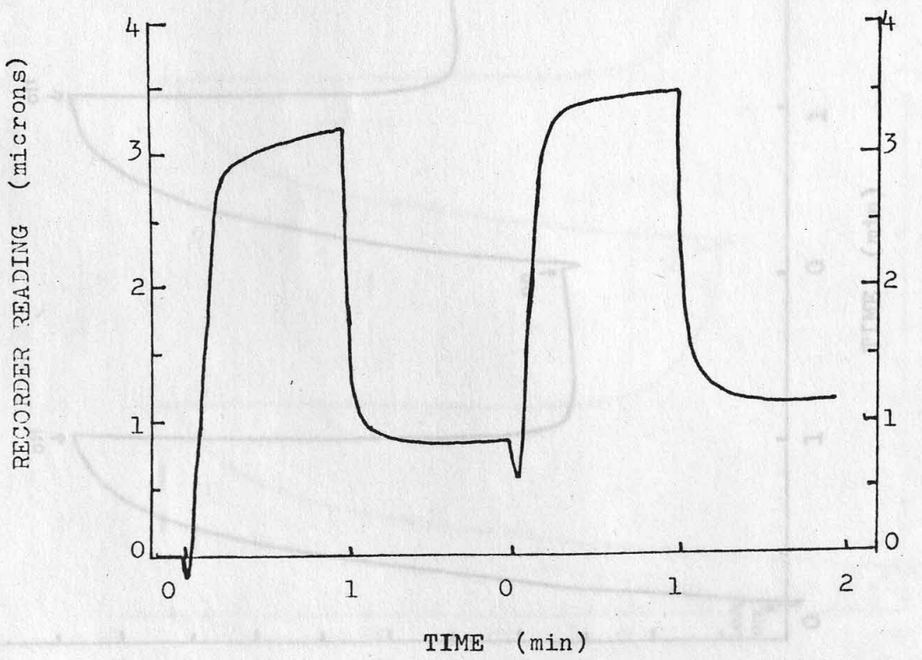


Figure V.9 Deformation properties of HEMA-VP(30:70) hydrogel under the eye-lid pressure.

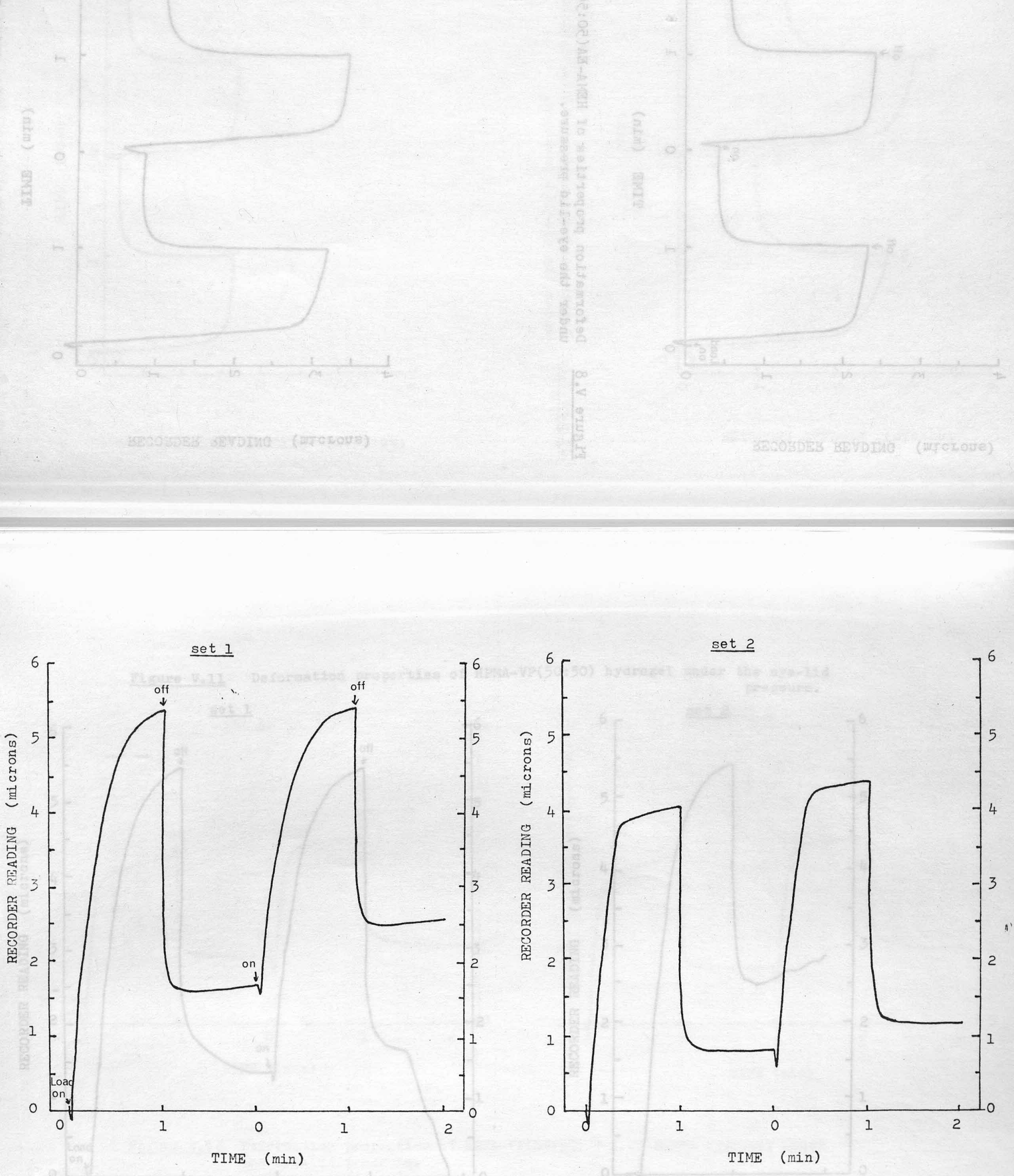
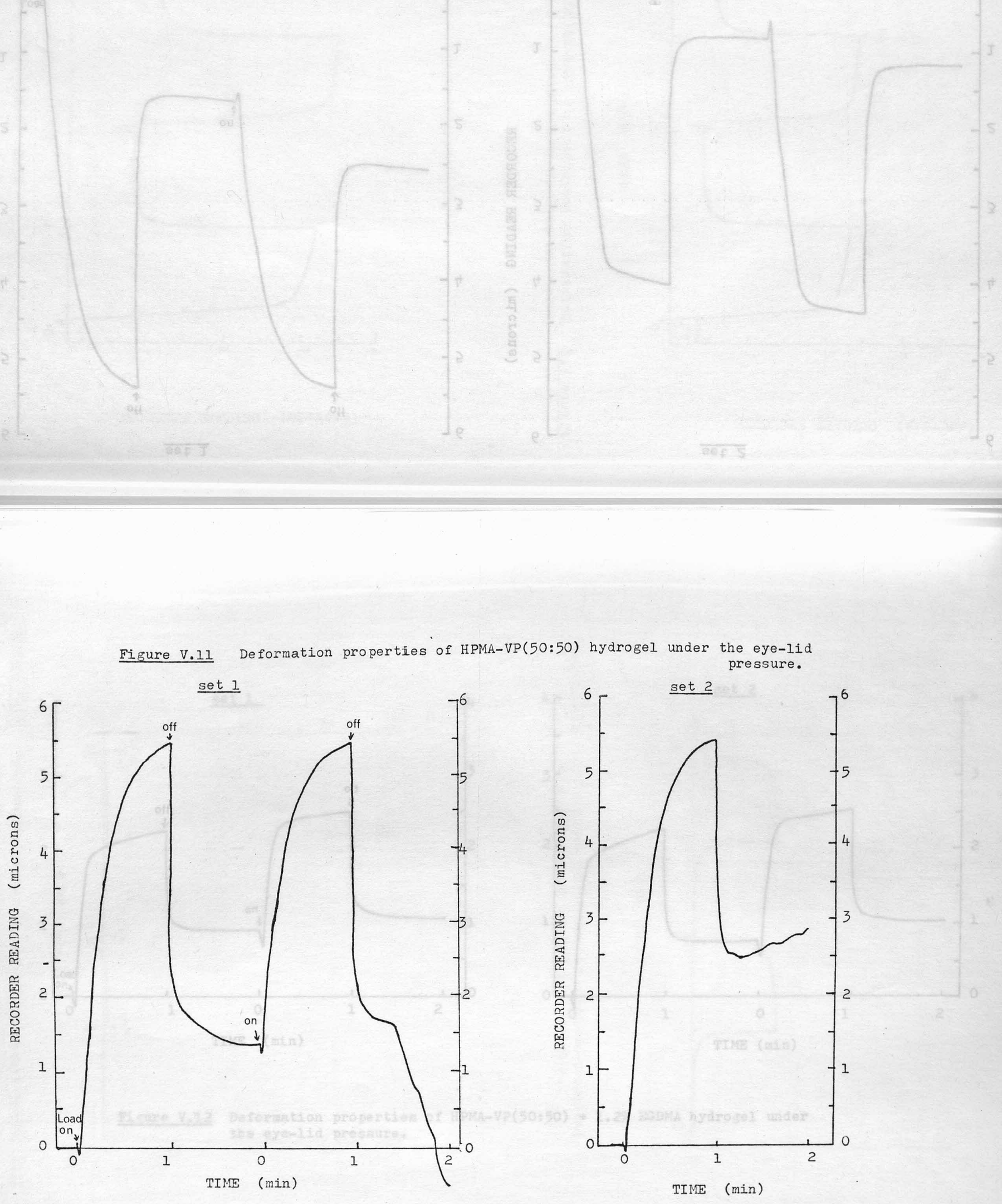


Figure V.10 Deformation properties of HEMA-VP(50:50) hydrogel under the eye-lid pressure.

Figure V.11 Deformation properties of HPMA-VP(50:50) hydrogel under the eye-lid pressure.



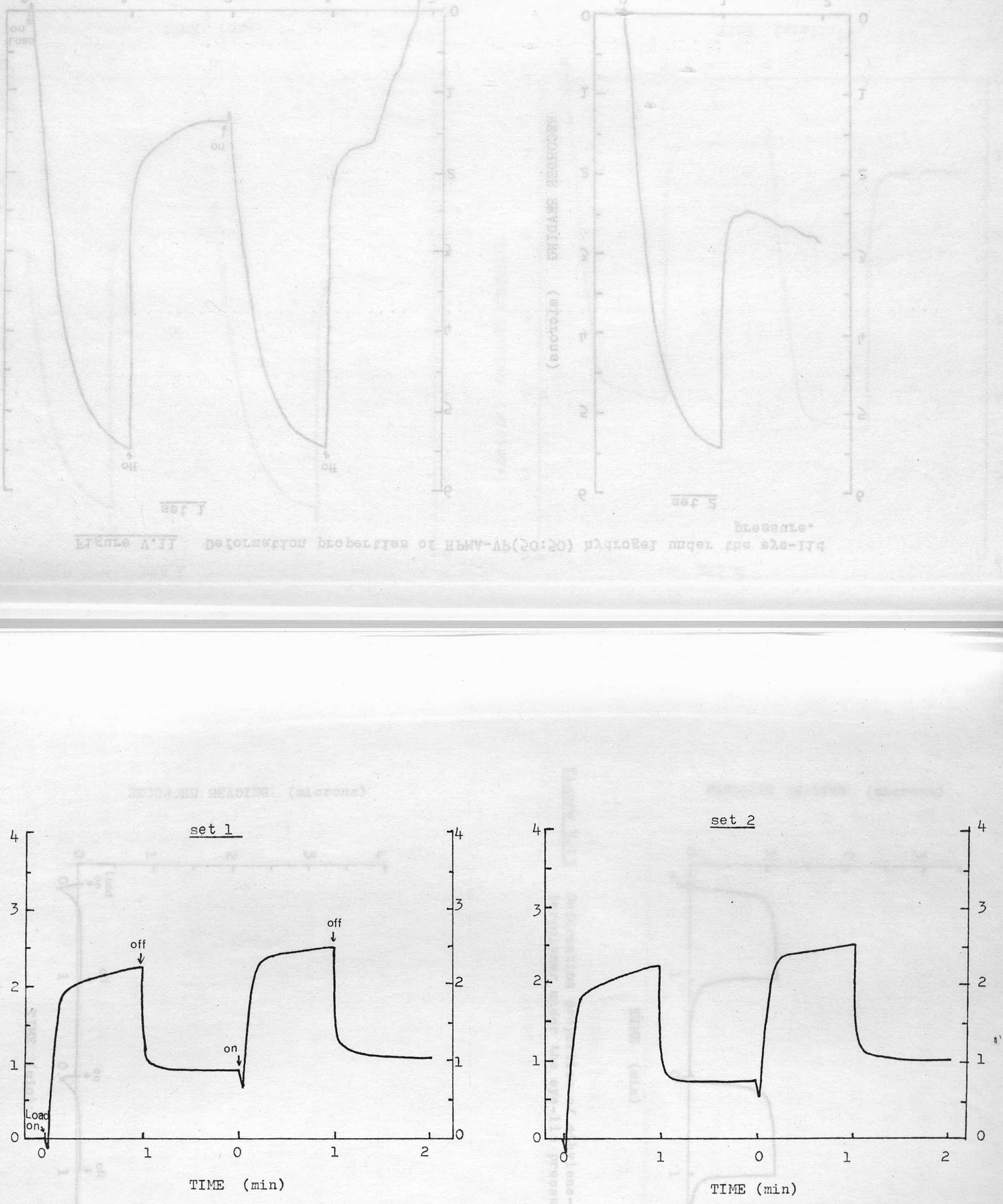


Figure V.12 Deformation properties of HPMA-VP(50:50) + 1.2% EGDMA hydrogel under the eye-lid pressure.

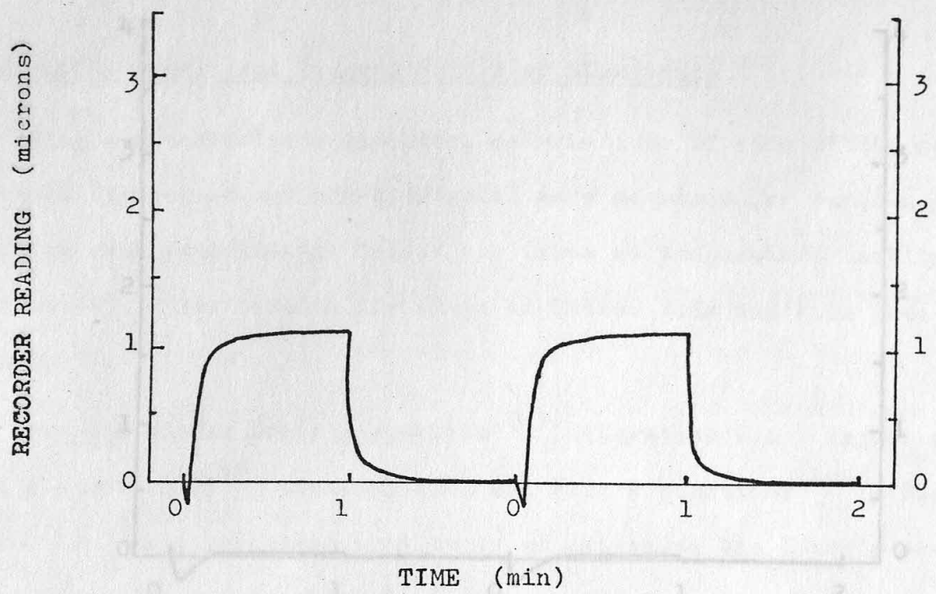


Figure V.13 Deformation properties of ethylene-propylene terpolymer under the eye-lid pressure.

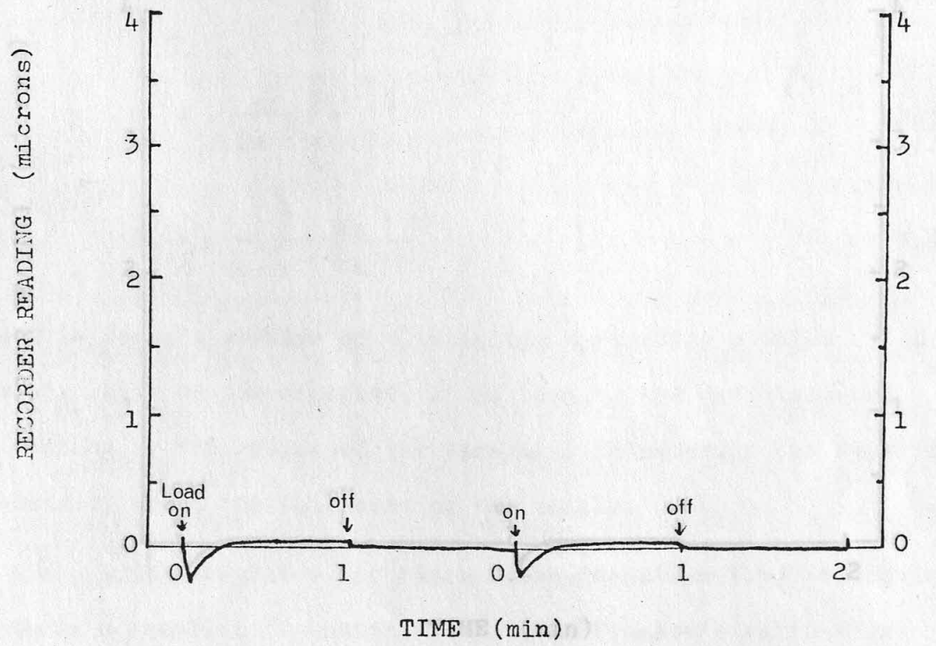


Figure V.14 Deformation properties of poly(4-methyl pent-1-ene) under the eye-lid pressure.

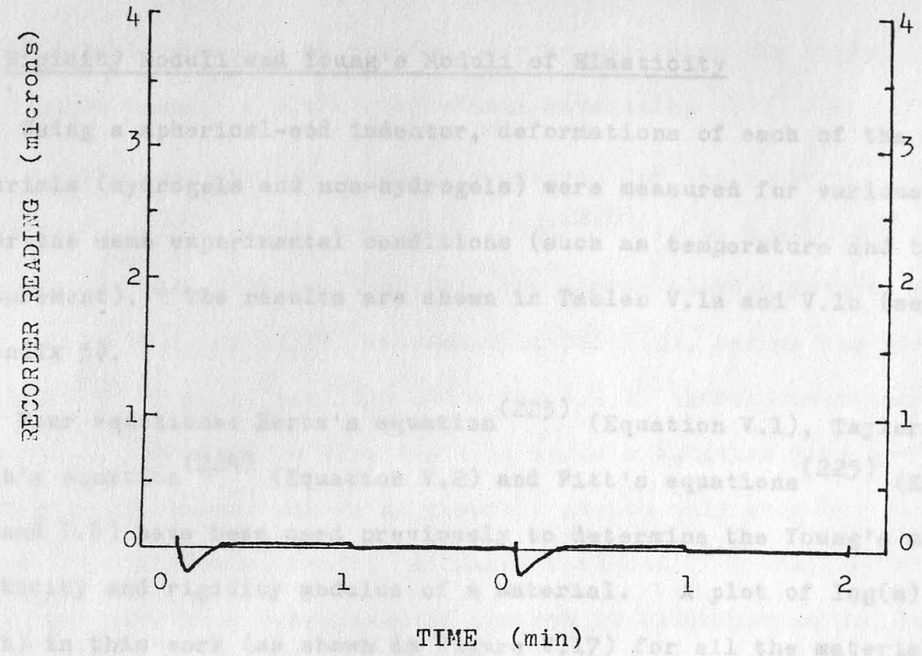


Figure V.15 Deformation properties of polystyrene under the eye-lid pressure.

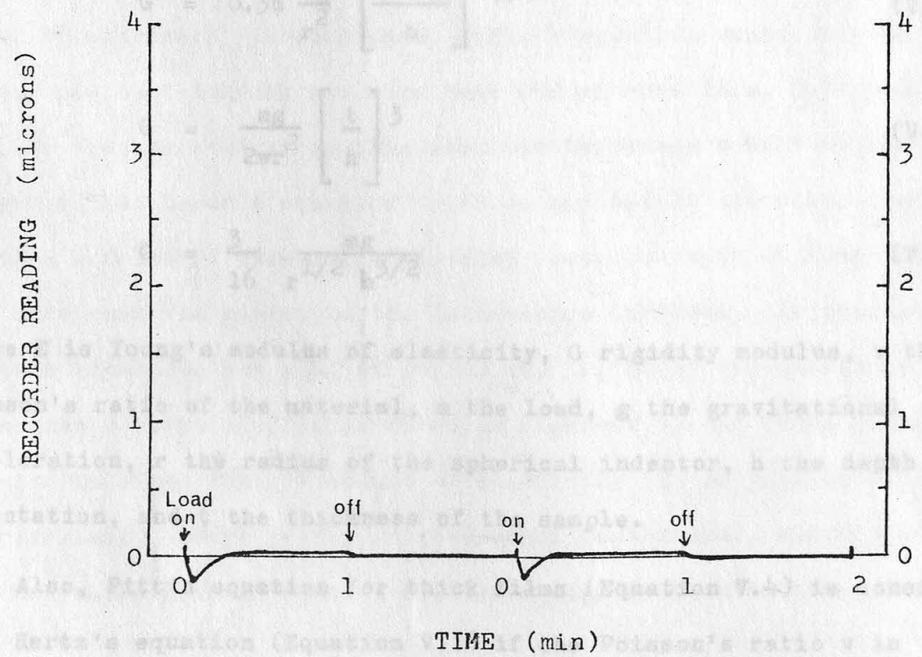
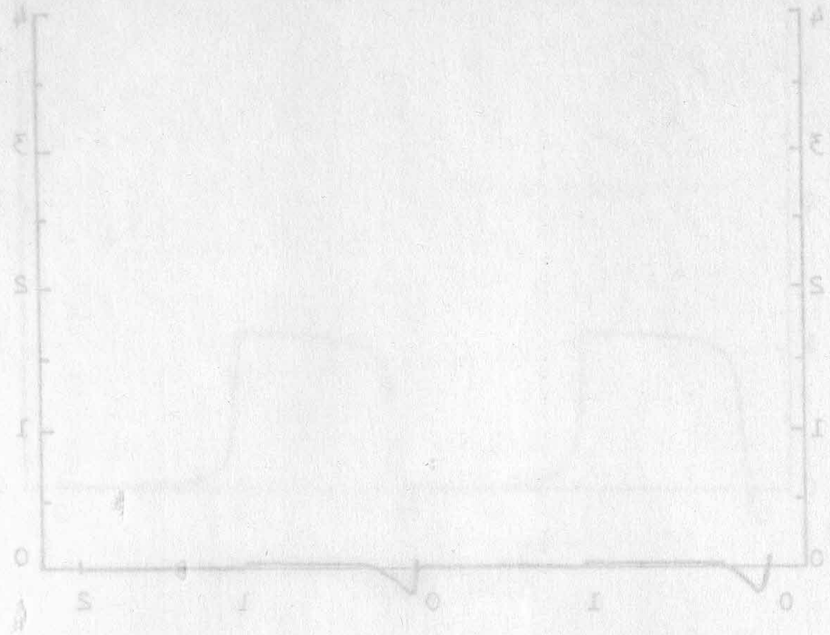


Figure V.16 Deformation properties of poly(methyl methacrylate) under the eye-lid pressure.





V.3.B. Rigidity Moduli and Young's Moduli of Elasticity

Using a spherical-end indentor, deformations of each of the several materials (hydrogels and non-hydrogels) were measured for various loads under the same experimental conditions (such as temperature and time of measurement). The results are shown in Tables V.1a and V.1b (see Appendix 5).

Four equations: Hertz's equation[223] (Equation V.1), Taylor and Kragh's equation[224] (Equation V.2) and Pitt's equations[225] (Equations V.3 and V.4) have been used previously to determine the Young's modulus of elasticity and rigidity modulus of a material. A plot of log(m) vs. log(h) in this work (as shown in Figure V.17) for all the materials studied gave a slope of 3/2, and so Pitt's equation for thin film (Equation V.3) would not apply here.

$$\frac{E}{1 - v^2} = \frac{3}{4} \frac{mg}{r^{1/2} h^{3/2}} \qquad (V.1)$$

$$G = 0.36 \frac{mg}{r^2} \left[\frac{t - h}{h}\right]^{3/2} \qquad (V.2)$$

$$G = \frac{mg}{2\pi r^2} \left[\frac{t}{h}\right]^3 \qquad (V.3)$$

$$G = \frac{3}{16} \frac{mg}{r^{1/2} h^{3/2}} \qquad (V.4)$$

where E is Young's modulus of elasticity, G rigidity modulus, v the Poisson's ratio of the material, m the load, g the gravitational acceleration, r the radius of the spherical indentor, h the depth of indentation, and t the thickness of the sample.

Also, Pitt's equation for thick films (Equation V.4) is consistent with Hertz's equation (Equation V.1) if the Poisson's ratio v in Equation V.1 is 0.5. Therefore only two equations (i.e. Equations V.1 and

and V.2) could possibly be applied here to determine the rigidity moduli and Young's moduli of elasticity of the materials.

Hertz's equation was originally derived for the contact of two perfectly elastic bodies, but later found[226] to be applicable to viscoelastic materials provided that the simple elastic modulus was replaced by the appropriate viscoelastic function, and so has been applied[227] to paint coatings using the I.C.I. Microindentation Apparatus. As shown in Equation V.1, Hertz's equation does not include the sample thickness, and so is strictly obeyed only when the sample is of infinite thickness, but it has been mentioned[228] that Hertz's equation could still be applied if the sample thickness is not less than five times the radius of circle of contact between the indenting sphere and the sample. It has been shown[229] that the radius (c) of the circle of contact could be related to the radius (r) of the indentor and the depth of indentation (h) by the equation $c = r^{½}.h^{½}$, which means that if r = 0.08 cm. (i.e. 1/16 inch diameter, which was used in this work), then an indentation of 1 micron (i.e. 10^{-4} cm.) would give a "c" value of about 28 microns. In such case, Hertz's equation could only be applied if the sample thickness was more than 140 microns (i.e. 0.14 mm). Since all the samples studied in this work had thickness > 0.14 mm, it was expected that Hertz's equation could be applied to calculate their Young's modulus of elasticity provided that the depth of indentation was not more than one micron in the deformation studies. As predicted by Hertz's equation, the plot of log(m) vs. log(h) of the materials studied here gave a slope of 3/2, as shown in Figure V.17 and Table V.2a. However only the Young's moduli of elasticity (E) of poly(methyl methacrylate), polystyrene, poly(4-methyl pent-1-ene), dehydrated PHEMA and dehyrated HEMA-St (90:10) copolymer were calculated using Hertz's

equation because they had indentations of not more than 1.5 microns within the range of loads studied, and the results are shown in Table V.2a. It was also possible to calculate the rigidity moduli (G) of these materials from the relation $E = 2(1+v)G$ by using the Poisson's ratio v of 0.35 and 0.32 for poly(methyl methacrylate) and polystyrene[230] respectively, and by assuming a similar value (say, 0.33) for poly(4-methyl pent-1-ene), dehydrated PHEMA and dehydrated HEMA-St (90:10). The rigidity moduli of these materials thus calculated are also shown in Table V.2a.

For the other materials such as EPT (i.e. ethylene propylene terpolymer, SBR (i.e. styrene butadiene rubber), PHEMA hydrogel and HEMA-St (90:10) hydrogel studied in this work, Hertz's equation was not expected to apply successfully as these materials gave a comparatively larger deformation (i.e. $>$ 1 micron) than what Hertz's equation could allow for at the sample thickness of about 200 microns (i.e. 0.2 mm) and with the spherical indentor of 0.160 cm. diameter. During the study of rigidity moduli of thin soft coatings, Taylor and Kragh[224] obtained an empirical equation (i.e. Equation V.2) which was valid for gelatin gels of rigidity moduli up to 1.5×10^5 dynes/cm^2 and rubbers of rigidity moduli 10^8 dynes/cm^2. This equation was obtained under the experimental conditions of spherical indentor radii between 0.08 and 0.32 cm., and of sample thicknesses betwen 75 and 800 microns, both of which covered the conditions used in this work (indentor radius 0.08 cm. and sample thickness $\sim$200 microns). Hence, Taylor and Kragh's equation which took sample thickness into consideration in the calculation, was used here to calculate the rigidity moduli of EPT, SBR, hydrated PHEMA and hydrated HEMA-St (90:10) materials, since a plot of log(m) vs. log(h/(t-h)) of these materials, as shown in Figure V.18, gave a slope of 3/2 as predicted

by the equation. The rigidity moduli, G, thus calculated are shown in Table V.2b. The Young's moduli of elasticity, E, of these materials could be calculated from $E = 2(1+v)G$, provided that their Poisson's ratios, v, were known. Although the v values of EPT, SBR, PHEMA hydrogel and HEMA-St (90:10) hydrogel were not actually known, they could by predicted here since it has been concluded[230] from many experimental facts that the Poisson's ratio, v, is 0.5 for rubbery materials and about 0.3 for glassy materials. If 0.5 was taken as the v value for the rubbery materials, EPT, SBR and hydrated PHEMA, and 0.4 for the less rubbery material HEMA-St (90:10) hydrogel, their Young's moduli of elasticity were calculated with an uncertainty of about $\pm$ 3%. The results are also shown in Table V.2b. A decrease from the order of 10^{10} to 10^{7} in both the rigidity modulus and Young's modulus of elasticity for PHEMA on hydration (equilibrium water content 36%), as shown in Tables V.2a and V.2b, is due to the presence of water in the hydrogel.

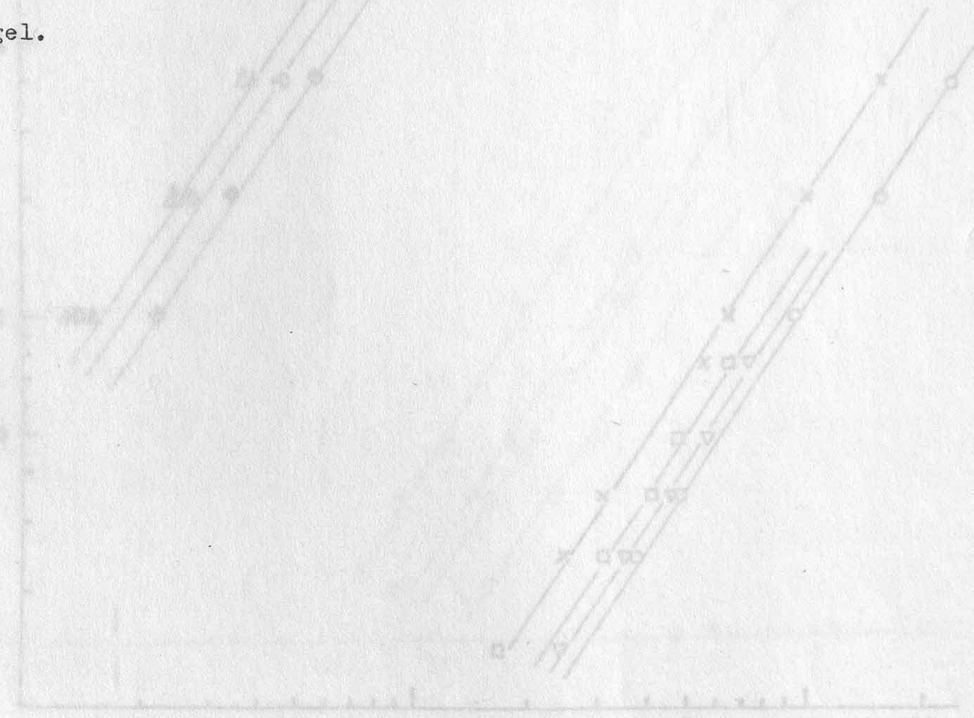


Figure V.17 The correlation of log(load) with log(indentation) for the various materials.

		Thickness (mm)
Δ	PMMA	0.18
D	Polystyrene	0.15
•	Dehydrated PHEMA	0.15
+	Dehydrated HEMA – St (90 : 10)	0.20
◆	Poly (4 – methyl pent-1- ene)	0.22
×	EPT	0.20
o	SBR	0.45
▽	PHEMA hydrogel	0.18
□	HEMA – St (90 : 10) hydrogel	0.22

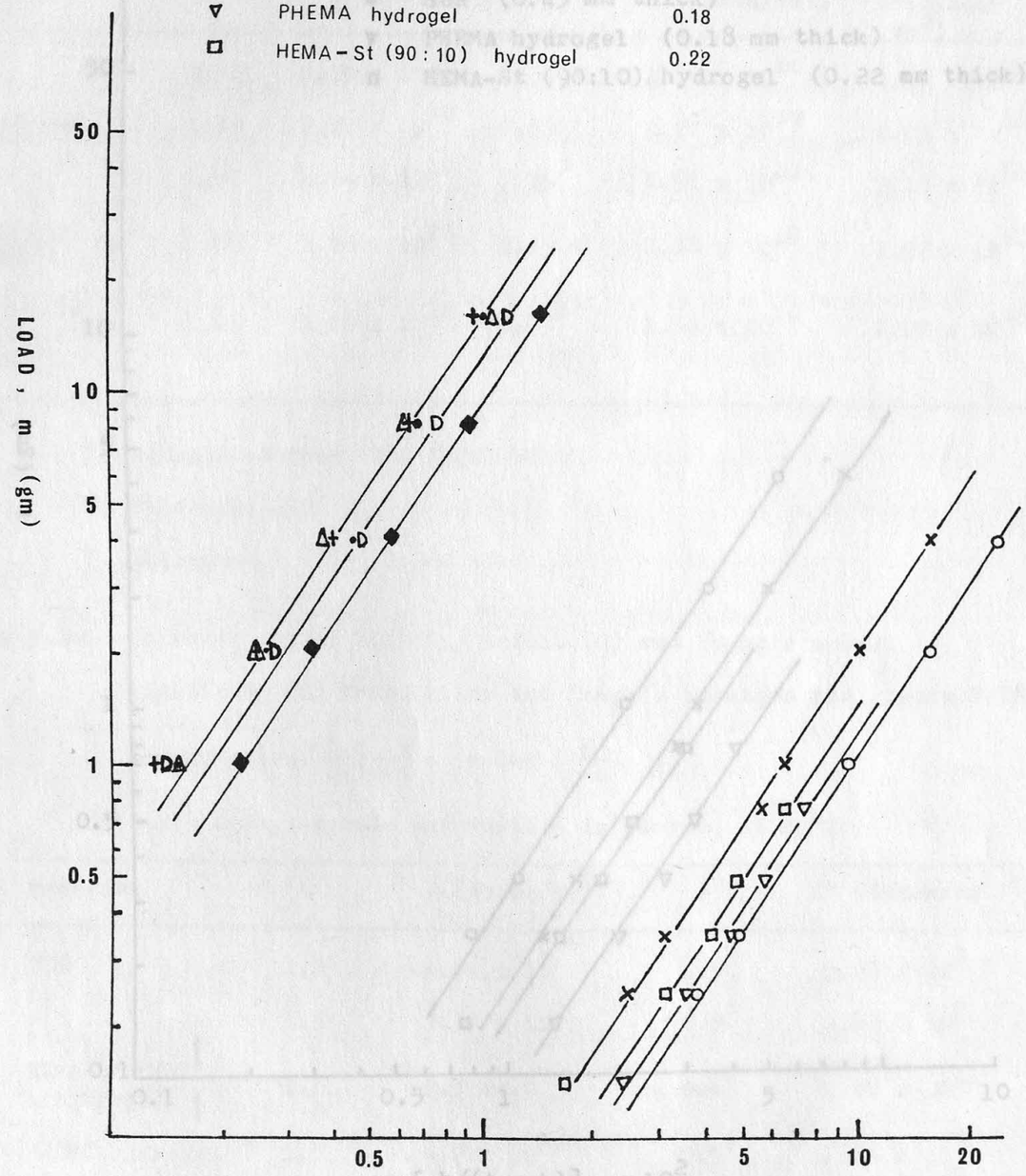


Figure V.18 The correlation of log(m) with log[h/(t - h)] for the various materials.

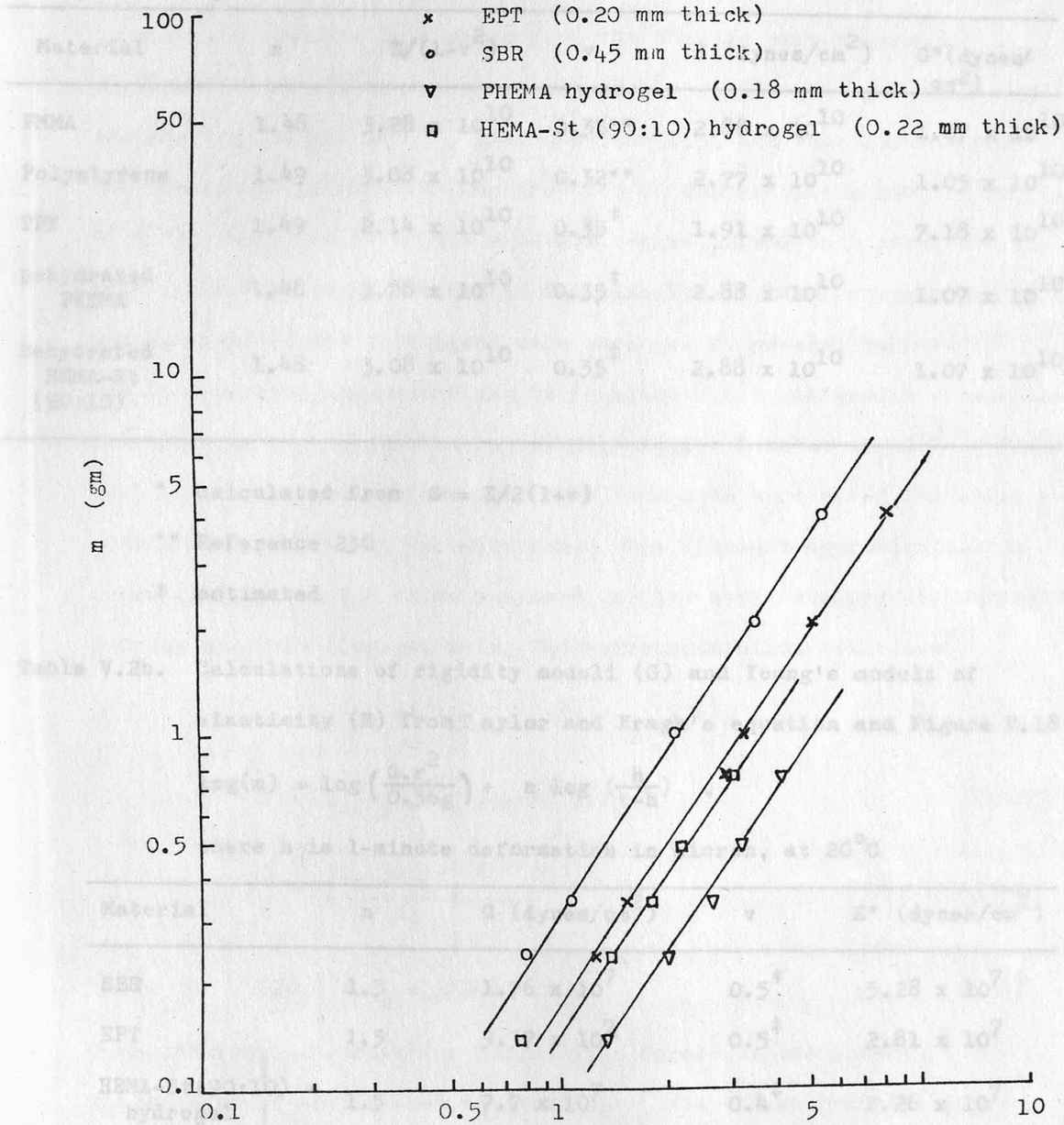


Table V.2a. Calculations of Young's moduli of elasticity (E) and rigidity moduli (G) from Hertz's equation and Figure V.17.

$$\log(m) = \log\left[\frac{4}{3g} \times \frac{E}{1-v^2} \times r^{\frac{1}{2}} \times \frac{1}{10^{4n}}\right] + n \log(h)$$

where h is 1-minute deformation in micron, at 20°C

Material	n	$E/(1-v^2)$	v	E (dynes/cm^2)	G*(dynes/cm^2)
PMMA	1.48	3.28×10^{10}	0.35**	2.88×10^{10}	1.07×10^{10}
Polystyrene	1.49	3.08×10^{10}	0.32**	2.77×10^{10}	1.05×10^{10}
TPX	1.49	2.14×10^{10}	0.33‡	1.91×10^{10}	7.18×10^{10}
Dehydrated PHEMA	1.48	3.28×10^{10}	0.35‡	2.88×10^{10}	1.07×10^{10}
Dehydrated HEMA-St (90:10)	1.48	3.08×10^{10}	0.35‡	2.88×10^{10}	1.07×10^{10}

* calculated from $G = E/2(1+v)$

** Reference 230

‡ estimated

Table V.2b. Calculations of rigidity moduli (G) and Young's moduli of elasticity (E) from Taylor and Kragh's equation and Figure V.18

$$\log(m) = \log\left(\frac{G.r^2}{0.36g}\right) + n \log\left(\frac{h}{t-h}\right) ,$$

where h is 1-minute deformation in micron, at 20°C

Material	n	G (dynes/cm^2)	v	E* (dynes/cm^2)
SBR	1.5	1.76×10^7	0.5‡	5.28×10^7
EPT	1.5	9.37×10^7	0.5‡	2.81×10^7
HEMA-St(90:10) hydrogel	1.5	7.7×10^7	0.4‡	2.26×10^7
PHEMA hydrogel	1.5	4.7×10^7	0.5‡	1.41×10^7

* calculated from $E = 2(1+v)G$

‡ estimated.

V.3.C. Determination of Cross-link Density of a Hydrogel

A preliminary investigation into determining the cross-link density of a hydrogel (PHEMA) was carried out using the I.C.I. Microindentation Apparatus.

It has been established that the cross-link density of a rubber vulcanisate could be determined from the tensile stress-strain measurements of either unswollen or swollen samples using Mooney, Rivlin and Sauders' equation[231], i.e. Equation V.5, and that for a highly swollen rubber network (volume fraction of rubber in the swollen gel $\approx$ 0.2), the C_2 value of the equation became effectively zero, and so equation V.5 could be written as Equation V.6. Later compression modulus measurement techniques were employed by several workers[232,233], who made various approximations to Equation V.6 to determine cross-link densities of highly swollen rubber networks. Stuckey et al[234] found that all the approximations that have been made were valid for benzene-swollen natural rubber vulcanisates. The simplest approximation is shown in Equation V.7 which was used in this work to study the cross-link density of PHEMA hydrogel using the microindentation technique

$$F = 2A_o.v_r^{-1/3}.(\lambda - \lambda^{-2}).(C_1 + \lambda^{-1}C_2) \qquad (V.5)$$

$$F = 2A_o.v_r^{-1/3}.(\lambda - \lambda^{-2}).C_1 \qquad (V.6)$$

$$\frac{F}{h} = \frac{2C_1.3A_o.v_r^{-1/3}}{h_o} \qquad (V.7)$$

where $(2C_1) = G_c = \nu RT$ (V.8)

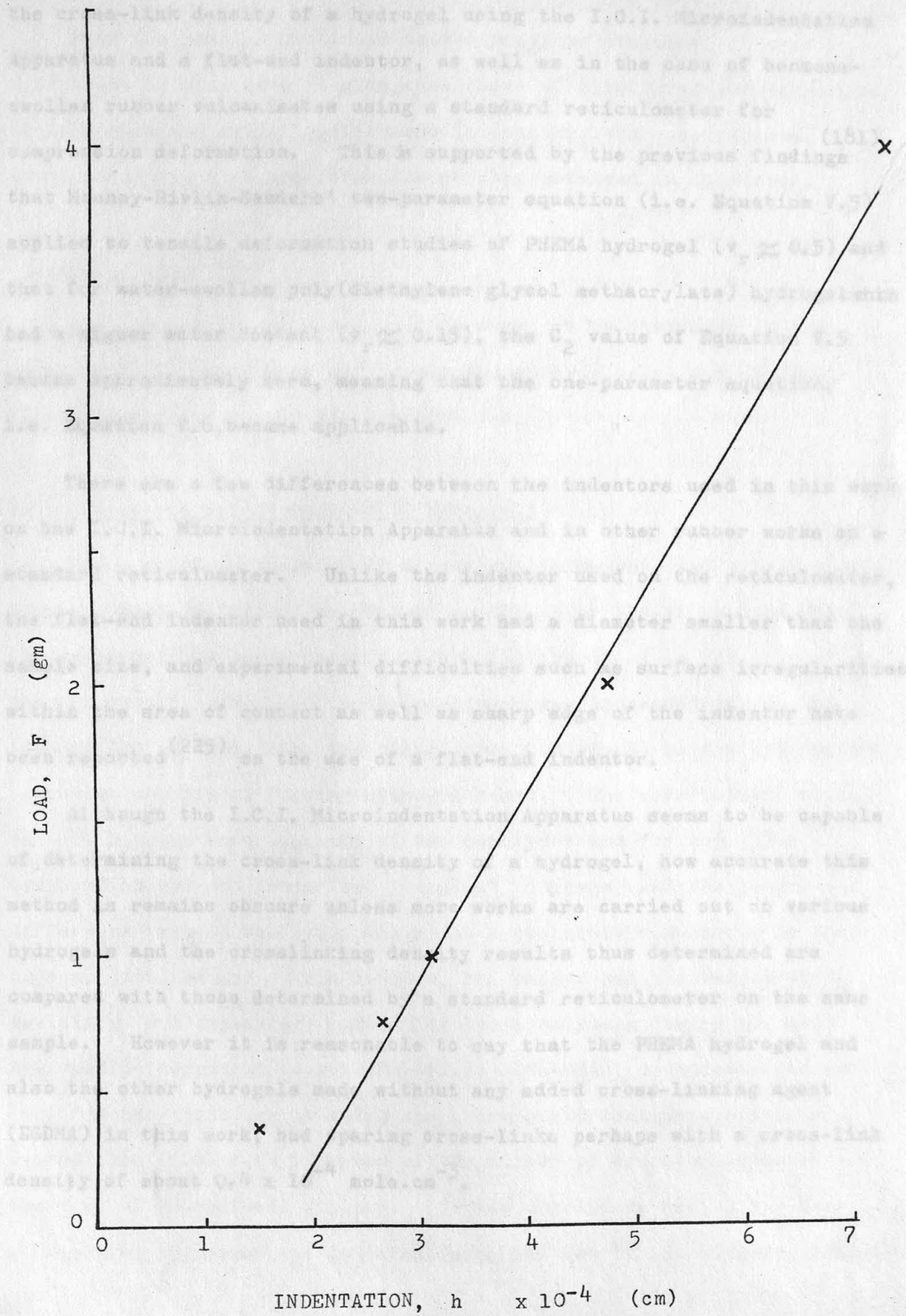
F is the tensile stress (or compression force) on the sample, A_o the area of sample when dehydrated, v_r the volume fraction of the polymer in the swollen gel, λ the relative extension, h_o the initial height of the sample when dehydrated, h the deformation of the swollen sample caused by the applied force, C_1 and C_2 are two constants

characterising the network, G_c the Gaussian modulus of the network, ν the cross-link density, R the gas constant, and T the test temperature in Kelvin.

Since Equation V.7 was found to be valid only for highly swollen rubber networks (with $v_r < 0.3$)[234], the application of this equation to determine cross-link density of PHEMA hydrogel ($v_r \approx 0.5$ in water) necessitates the PHEMA network to be swollen in a better solvent. Therefore PHEMA in this work was swollen with dimethyl sulphoxide which gave a v_r value of 0.171, and the deformation studies were made on the dimethyl sulphoxide-swollen PHEMA sample.

Using a flat-end indentor, deformations of the swollen PHEMA sample were measured for various loads, and the results are shown in Table V.3. A plot of compression load F (in gm) vs. depth of deformation (in micron) gave a slope of 3/4.5, as shown in Figure V.19. Using Equation V.7 and Equation V.8 with $v_r = 0.171$, $A_o = 3.84 \times 10^{-3} cm^2$ and $h_o = 120$ microns, the Gaussian modulus (G_c) was calculated as 8.3×10^2 gm/cm^2, and hence the effective network density of the gel 3.5×10^{-5} mole/cm^3. The effective network density in PHEMA hydrogel has been assumed[182], from the study of equilibrium deformations of PHEMA hydrogel using tensile-stress-strain measurements, to be attributed by chemical cross-links and entanglements. This means that although the effective network density of PHEMA gel was found to be 0.35×10^{-4} mole/cm^3, the actual chemical cross-links should be less than this value, and since the cross-linking efficiency in bulk polymerisation of HEMA has been found[182] to be about 1, the amount of residual cross-linking agent (EGDMA) present in the monomer should be less than 0.17×10^{-4} mole/cm^3 or about 0.4% (w/w). This is quite consistent with the value, estimated from the G.L.C. analysis, of not more than 1% of residual EGDMA in the monomer HEMA.

Figure V.19 The plot of load (F) vs. indentation (h) for the dimethylformamide-swollen PHEMA gel.



So Equation V.7 and Equation V.8 seem to be applicable to determine the cross-link density of a hydrogel using the I.C.I. Microindentation Apparatus and a flat-end indentor, as well as in the case of benzene-swollen rubber vulcanisates using a standard reticulometer for compression deformation. This is supported by the previous findings (181) that Mooney-Rivlin-Sauders' two-parameter equation (i.e. Equation V.5) applied to tensile deformation studies of PHEMA hydrogel ($v_r \approx 0.5$) and that for water-swollen poly(diethylene glycol methacrylate) hydrogel which had a higher water content ($v_r \approx 0.15$), the C_2 value of Equation V.5 became approximately zero, meaning that the one-parameter equation, i.e. Equation V.6, became applicable.

There are a few differences between the indentors used in this work on the I.C.I. Microindentation Apparatus and in other rubber works on a standard reticulometer. Unlike the indentor used on the reticulometer, the flat-end indentor used in this work had a diameter smaller than the sample size, and experimental difficulties such as surface irregularities within the area of contact as well as sharp edge of the indentor have been reported(225) on the use of a flat-end indentor.

Although the I.C.I. Microindentation Apparatus seems to be capable of determining the cross-link density of a hydrogel, how accurate this method is remains obscure unless more works are carried out on various hydrogels and the crosslinking density results thus determined are compared with those determined by a standard reticulometer on the same sample. However it is reasonable to say that the PHEMA hydrogel and also the other hydrogels made without any added cross-linking agent (EGDMA) in this work, had sparing cross-links perhaps with a cross-link density of about 0.4×10^{-4} mole.cm^{-3}.

Chapter VI. OVERALL DISCUSSIONS AND CONCLUSIONS

From the results mentioned in the previous chapters, an overall discussion is made here to give some views of structures and properties of hydrogels and their significance in contact lens applications. Conclusions are also made from the results obtained in this work.

VI.1. Overall Discussions

All the results obtained in this work can be discussed in two sections: (A) structures and properties of hydrogels, and (B) their significance in contact lens applications.

VI.1.A. Structures and Properties of Hydrogels

For a copolymer made up of a hydrophilic monomer and a hydrophobic monomer (such as HEMA-St or HPMA-St or HPA-St or HEA-St or HEMA-MMA) the equilibrium water content on hydration increased as the content of the hydrophilic component in the copolymer was increased due to the interaction of water with the polar group (OH group) in the hydrophilic component capable of forming hydrogen bonds. The correlation between the equilibrium water content of the copolymer and the copolymer composition was not linear but "concave" in shape, and the larger the difference between the hydrophilic and hydrophobic components in their hydrophilicities and steric factors, the larger was the degree of deviation from linearity. Thus for two copolymers having the same hydrophilic component (e.g. HEMA-St and HEMA-MMA), such deviation was less for the copolymer in which the hydrophobic component was less hydrophobic (i.e. both in terms of the number of hydrocarbon units and the bulk of substituent groups). If two copolymers having the same hydrophobic component are compared (e.g. any two of the HEMA-St, HPMA-St,

HEA-St and HPA-St copolymers), this deviation was less for the copolymer in which the hydrophilic component was more hydrophilic (i.e. had a smaller number of hydrocarbon groups and less bulky groups). Hence the incorporation of a hydrophobic component into the hydrophilic polymer resulted in not only a dilution of the hydrophilic component, but also a modification of the environment of the polar group in the hydrophilic component for water absorption. The steric hindrance around the polar group seems to be the predominant factor in restricting the water absorption of the copolymer if both the polar group and the solvent (water) are fixed. Similar arguments could also be applied to copolymers containing two hydrophilic components (e.g. HEMA-VP and HPMA-VP), and to terpolymers having two hydrophilic components and one hydrophobic component (e.g. HEMA-VP-St, HPA-VP-St and HEMA-HPA-St).

The presence of chemical cross-links (covalent bonds) in a copolymer or a terpolymer (by incorporating a cross-linking agent, e.g. EGDMA) lowered its water absorption, and the equilibrium water content-copolymer component relationship was expected to deviate more from linearity. It is understood that chemical cross-links restricted the rotation of polymer chains and reduced maximum exposure of the polar groups to water.

The insolubilities of the copolymers and terpolymers were attributed to chemical cross-links and hydrophobic bondings. Although chemical cross-links gave rise to the primary structure and limited the solubility of a hydrophilic polymer in water, hydrophobic bondings between non-polar groups in the polymer could also reduce its equilibrium water content as they acted as "cross-links". Hydrophobic bondings result from the entropically unfavoured solvation of non-polar groups by water, and could be important in hydrogels particularly with low degree of chemical cross-links. This is seen in the effects of

hydration temperature and sodium chloride (and other microsolutes[119]) on the equilibrium water contents of hydrogels. Although hydrogen bondings between polar groups have been attributed primarily to the stability of some native protein in water[208], it is unlikely that such hydrogen bondings are also present in the hydrogels studied here. It is more probable that hydrogen bondings were present in those hydrophilic polymers in the non-gel state.

A hydrogel is defined in this thesis as a cross-linked hydrophilic polymer having a glass transition temperature in the hydrated state lower than the room temperature, with the perturbation of the polymer network caused by the plasticisation of water. Not all the copolymers containing a hydrophilic and a hydrophobic component studied in this work were hydrogels on hydration. The copolymer compositions at which they became hydrogels were found by density measurements and flexibility assessments of the hydrated materials. Thus on hydration, HEMA-St copolymer became a hydrogel if it contained more than about 75% HEMA, HPA-St copolymer more than about 60% HPA, HEA-St copolymer more than about 60% HEA, and HEMA-MMA copolymer more than about 40% HEMA. The transition from non-hydrogels to hydrogels seemed to be gradual rather than abrupt, but to different degrees for the various copolymers. At approximately these copolymer compositions, there was an onset of perturbation of the polymer matrix with swelling of the network. It could be deduced, from the equilibrium water content-copolymer composition correlations, that these copolymers, HEMA-St, HPA-St, HEA-St and HEMA-MMA, became hydrogels if their equilibrium water contents were more than approximately 10%, 9%, 9% and 8% (on weight basis) respectively. These corresponded to approximately one mole of water for each mole of the hydrophilic group in each of the copolymers. It has been found[221] that mechanical properties of PHEMA changed

abruptly at a water content of 10%, and again this corresponded to about one mole of water for each mole of HEMA. Baresova[179,235], in the studies of density and relaxation properties of PHEMA hydrogel at low temperatures, came to the conclusion that about 0.5 molar concentration (i.e. water content about 11% on our basis) of water was more strongly bound to PHEMA and formed a monolayer even at low temperature. If this could be applied similarly to the copolymers studied in this work, then water bound more strongly to the HEMA-St, HPA-St, HEA-St and HEMA-MMA copolymers would be about 10%, 9%, 9% and 8% respectively.

The results that "dissolved" oxygen permeability coefficients of hydrogels increased exponentially with their equilibrium water contents over a wide range of hydration (at least between 20% and 80%) indicate that oxygen diffused through the water phase in the hydrogels. This is understandable because water fills up the interstices of the polymer network in a hydrogel. Due to the plasticisation effect of water, the polymer chains in a hydrogel have some degrees of mobility. It is probable that thermal fluctuations of both the polymer chains and water molecules result in free volumes or "holes" (not fixed pores, but fluctuating with time) for oxygen molecules to "jump" from one "hole" to another in the direction of the oxygen concentration difference. Increasing temperature increased mobilities of polymer chains and water molecules and hence increased "dissolved" oxygen permeabilities of hydrogels, but on the other hand, increasing temperature reduced equilibrium water contents of hydrogels in some cases.

There have been some revelant works regarding states of water in hydrophilic polymers. Jadwin et al[111] have divided the water present in PHEMA hydrogel into water bound to the hydrophilic OH groups of PHEMA ("primary" water) and unbound water ("secondary" water). By assuming that sodium chloride dissolved only in the "secondary"water, they have estimated that the "primary" water in PHEMA hydrogel (with an equilibrium water content of 38%) is about 25% and "secondary" water

about 13%. This means that there were about three "primary" water molecules for each OH group of PHEMA. The validity of their assumption is not clear because sodium chloride may interact with the polymer in the hydrogel. Water present as "solid-like" (7%) and "bound" (13%) have also been reported by Tait et al[236] on starch at low moisture content (20%). From the D.S.C. and N.M.R. studies of poly(glycerol methacrylate) hydrogel [i.e. poly(1,2-dihydroxypropyl methacrylate)], Yasuda et al[237] came to the conclusion that (1) about 26% water content of poly(glycerol methacrylate) hydrogel was still bound to the polymer ("bound" water) at low temperatures, while the rest of the water was freezable into ice ("free" water), and (2) the "bound" water though highly associated with the polymer, still remained substantial mobility. Two states of water ("primary" and "secondary" waters) have also been determined in homogeneous cellulose acetate membranes by water diffusion techniques[238].

The oxygen permeability results together with density results obtained in this work seemed to suggest three possible states of water in a hydrogel depending on the degree of hydration. While various names have been used to describe states of water in a hydrophilic polymer, as shown above, it is decided in this thesis to use the terms "primary" water and "secondary" water, and in addition, "liquid-like" water. If the "primary" water (i.e. hydrogen-bonded to the polymer) in the copolymers studied in this work was not more than about 10% water content (from the density results, discussed earlier), then any additional water in the hydrogel would exist as "secondary" water which possibly started at the region of the onset of polymer swelling (i.e. when the polymer became a hydrogel on hydration). When the equilibrium water content of a hydrogel was about 40% to 50%, "liquid-

like" water seemed to start to exist in addition to the "primary" and "secondary" waters, as suggested by the "dissolved" oxygen permeability results.

As shown in Figure IV.14 for the correlation of "dissolved" oxygen permeability coefficients and equilibrium water contents of hydrogels, the "dissolved" oxygen permeability coefficients of the hydrated copolymers in the region of "primary" water (i.e. less than about 10% equilibrium water content) decreased sharply and then increased slightly, suggesting that an increase in HEMA content in this region of compositions of HEMA-St copolymers resulted in an increase in the packing of the polymer chains, and also that water molecules might be in a "cross-linking" state in which a water molecule was bridged between polar groups of the polymer chains and had a very low mobility. A sharp drop in the oxygen permeability of a non-polar polymer due to incorporation of a polar component has also been reported in the case of butadiene-acrylonitrile system[239]. The concept of "cross-linking" water molecules has been proposed and used to explain the maximum viscosity observed in polyethylene glycol dissolved in a small amount of water[240], and also to explain an increase in shear modulus of poly(vinyl alcohol) film during the initial period of addition of water to the film[241]. But as the "primary" water was increased to near one molecule of water for each molecule of the polar group of the polymer chain (i.e. near the onset of the swelling of the polymer network), the possibility of having "cross-linking" water molecules became less and also the interaction between polar groups of the polymer chains was shielded by a monolayer of water molecules, resulting in the flexibility of the polymer chains and hence a slight increase in "dissolved" oxygen permeabilities of the hydrated polymers in the later

part of the "primary" water region. As "secondary" water existed due to an increase in equilibrium water content, the thermal fluctuations of the "secondary" water molecules and the plasticised polymer chains increased, resulting in an increase in "dissolved" oxygen permeabilities. The intrinsic flexibility of the polymer chains (i.e. without water plasticisation) contributed to the oxygen permeabilities of the hydrated copolymers in these two regions of water ("primary" and "secondary"), and so the "dissolved" oxygen permeability coefficient of a hydrogel was affected by its chemical structure as well as its equilibrium water content, as observed in the higher "dissolved" oxygen permeability coefficients of hydrogels of some HEMA-methacrylate copolymers and HEMA-acrylate copolymers compared with that of HEMA-St hydrogel at the same equilibrium water content. As the equilibrium water content was further increased to about 40% - 50%, the third state of water ("liquid-like" water) started to exist, as observed in a big increase in the "dissolved" oxygen permeability coefficient in this region of equilibrium water content. The high thermal fluctuations of "liquid-like" water molecules contributed predominantly to the high oxygen permeability of the hydrogel, and in fact, the oxygen permeability coefficient curve passed through 100% bulk water. Opaque PHEMA hydrogel with equilibrium water content about 60%, which is thought to contain clusters of water, has been reported to have a "dissolved" oxygen permeability coefficient approaching that of bulk water[131]. Apparently the chemical structure of a polymer played only a small part in the oxygen permeability after about 50% in equilibrium water content.

The oxygen permeability results suggest that there were no distinct boundaries between "primary" water and "secondary" water, and between "secondary" water and "liquid-like" water, and in fact, the transition

from the "primary" water to "secondary" water and then to "liquid-like" water seemed to be gradual rather than abrupt. The oxygen permeability results indicate that the "primary", "secondary" and "liquid-like" waters in the hydrogels studied in this work were in the order of increasing mobility or decreasing water structure. There have been reports[242] that some ions are water structure promotors while others are structure breakers, and that polar groups such as OH and COOH destruct water structure and facilitate water transport through membranes. This may not be the exact situation in the hydrogels studied in this work which contain polar hydroxyl groups or amide groups, because of the presence of non-polar groups in the polymers, which are known[207a] to be water structure promotors. Although the real structure of water in the hydrogels is not clear, the net effect of the hydrophilic hydration and hydrophobic hydration seemed to be structuring of water in the hydrogels, as suggested by the oxygen permeability results.

The deformation properties of hydrogels, as measured with the I.C.I. Micro-indentation Apparatus were found to be dependent on the chemical structures and equilibrium water contents of the hydrogels. At the same equilibrium water content, a hydrogel containing a more rigid polymer chain had a smaller but more viscoelastic deformation under a load than a hydrogel containing a more flexible chain, and also the former showed a more retarded recovery than the latter. High water content hydrogels gave a larger deformation than lower water content hydrogels, but the nature of deformation and recovery depended on the equilibrium water content. Increasing cross-link density reduced the deformation of a hydrogel under a load and improved its elastic deformation and recovery properties.

The molecular dispersion of water in a hydrogel not only affected the copolymer composition at which the copolymer became a hydrogel on

hydration (compare HEMA-St and HEMA-MMA copolymers), but also affected the optical property of a hydrogel. A hydrogel was observed to be translucent or opaque if the copolymer from which the hydrogel was made was originally heterogeneous or if the hydration of a homogeneous copolymer caused phase separation. The former situation could arise if polymerisation of two or more monomers resulted in polymer chains of very much difference in solubilities (different polarities), possibly different monomer sequence and distribution in the polymer chain, which was a consequence of very much difference in reactivity ratio of the monomers (e.g. vinyl pyrrolidone-styrene copolymers). Hydration of a homogeneous copolymer caused phase separation if water was present as clusters in the hydrogel rather than molecularly dispersed in the whole network. Temperature induced translucency or opacity was transient and could be due to slow relaxation of the polymer chains in removing micro-separated water clusters from a swollen hydrogel.

VI.1.B. Significance of the Properties in Contact Lens Applications

All the results obtained in this work suggest that high water content hydrogels ($>$ 40% in equilibrium water content) are favoured to low water content hydrogels in oxygen permeability, density and refractive index, but unfavoured in deformation properties. High water content hydrogels have density approaching 1 (density of water) and refractive index approaching 1.35 (refractive index of the tear fluid). Although these two properties are not as critical as the oxygen permeability and deformation properties of a hydrogel in contact lens applications, still it has been suggested[135] that the density of an ideal contact lens should not be high (preferably near 1). Perhaps this is because a lighter contact lens may centre better at the cornea than a

heavier one. One possible advantage of having a hydrogel contact lens with a refractive index approaching that of the tear fluid is that a steeper curve in the contact lens is thus allowed, and this provides a wider range of lens curvatures in contact lens design.

A higher water content hydrogel gives a higher oxygen permeability, but how high the oxygen permeability coefficient of a hydrogel contact lens for a particular thickness is necessary for a continuous wear seems to be uncertain because of several factors affecting the actual function of a contact lens in the eye. If it is correct that for a cornea to show no sign of edema, a minimum oxygen tension of 11-19 mm.Hg. must be maintained at the epithelial surface of the cornea[64], i.e. a minimum oxygen flux of 3.5 $\mu l.cm^{-2}.hr^{-1}$ must transmit through a tight contact lens and enter the epithelium for its metabolic requirement[61], then the contact lens material for continuous wear should, for a 0.1 mm. thick lens, have a minimum "dissolved" oxygen permeability coefficient of 230 x 10^{-10} cc(S.T.P.).mm.cm^{-2}.sec^{-1}.cm Hg^{-1} at 34°C, i.e. a minimum equilibrium water content of about 52% (in 0.9% saline) at 34°C (normal eye temperature). If the required oxygen flux of 7 $\mu l.cm^{-2}.hr^{-1}$ found by Hill et al[54] is correct, then a 0.1 mm. thick continuous-wear contact lens should have a minimum "dissolved" oxygen permeability coefficient of 460 x 10^{-10} cc(S.T.P.)mm. cm^{-2}.sec^{-1}. $cm.Hg^{-1}$ at 34°C, i.e. a minimum equilibrium water content of 68% (in 0.9% saline) at 34°C. In fact, the oxygen flux into the cornea under <u>normal</u> conditions was found by Hill et al[54] as 4.8 to 7.2 $\mu l.cm^{-2}.hr^{-1}$. It is possible that the oxygen flux required by the cornea could go down to 3.5 $\mu l.cm^{-2}.hr^{-1}$ before corneal edema occurs. It is important in the contact lens design that the contact lens thickness should be mentioned with the "dissolved" oxygen permeability coefficient of a contact lens material, because a thicker contact lens will require a material of

higher "dissolved" oxygen permeability coefficient. In fact, this is a linear correlation, as discussed in Section IV.3.C. The conclusion made on the required "dissolved" oxygen permeability coefficient of a continuous-wear contact lens material could, however, be complicated by some factors. (1) the pumping action of the contact lens during blinking gives rise to tear circulation under the lens and hence increases the oxygen tension under the lens. (2) The tear fluid under the lens may circulate slightly during sleep; although blinkings by the eye-lids are unlikely to occur during normal sleep, the movement of the eye ball may be possible. (3) The oxygen consumption rate of a human cornea varies from one person to another.

With regard to carbon dioxide gas (the metabolic product) it has been reported that a high-water-content hydrogel of polyelectrolyte complexes has a carbon dioxide permeability coefficient 20 times of its oxygen permeability coefficient, whereas silicone rubber has only 5 times. (219)

That a higher water content hydrogel gives a larger deformation under a load than a low water content hydrogel leads to a prediction that a higher-water-content hydrogel contact lens will give a poorer visual acuity than a low-water-content hydrogel contact lens during wear. While some low-water-content hydrogel materials, e.g. HEMA-St (90:10) polymer, showed viscoelastic properties under a load and upon removal of the load, they might be more useful for daily-wear contact lenses than continuous-wear contact lenses because, in daily wear situations, shorter duration of loads are encountered and greater tear strength is required. In fact, contact lenses made of HEMA-St (90:10) hydrogel have been shown clinically[243] to give better visual acuity during wear than PHEMA hydrogel contact lenses, but give a minimum level of comfort in long term contact lens wear. One possible remedy to a high-water-content hydrogel is to control its cross-link density to give a better

dimensional stability and more elastic properties while still having a sufficient oxygen transmission for the corneal requirements, so that although the deformation of the high-water-content hydrogel contact lens is relatively large under the eye-lid/pressure (compared with a low-water-content hydrogel), the visual acuity of the contact lens during wear could possibly be improved by the "springiness" of the lens and also by the design and good fitting of the contact lens, which have been believed to play a part in affecting visual acuity during the contact lens wear.

Since it has been clinically proved that a commercial PHEMA hydrogel contact lens is wettable by the tear fluid, it is expected that high-water-content hydrogel contact lenses (equilibrium water content $\geq$ 40%) should be more wettable in the eye. Also high-water-content hydrogel contact lenses are expected to be softer and more biocompatible with the eye, and hence more comfortable to wear, than the existing PHEMA hydrogel contact lenses. Since very-high-water-content hydrogels have poor strengths and a large deformation under a load, it is logical to think that the appropriate equilibrium water content of a continuous-wear contact lens should be somewhere between 50% to 70%.

A possibility of using a low-water-content hydrogel (equilibrium water content 20% to 40%) for a continuous-wear contact lens is to use a very thin lens if it could be made. It is estimated in this work that for a 0.1 mm thick contact lens for continuous wear, a minimum "dissolved" oxygen permeability coefficient of 230×10^{-10} cc(S.T.P.). $mm.cm^{-2}.sec^{-1}.\ cm\ Hg^{-1}$ at 34°C is required, meaning that the equilibrium water content of the material in 0.9% saline at 34°C should be at least 52% (i.e. linear swell about 140%). Thus it can be foreseen that a perfect lathe-cut or compression-moulded contact lens in its dehydrated state will require highly skilled techniques and sophisticated equipments in the contact lens manufacturing process for a wide range of

contact lens design. Also, since the equilibrium water content of a hydrogel contact lens depends on several experimental conditions, accurate controls of monomer purity, monomer feed ratio, cross-linking agent concentration, polymerisation conditions, etc. are certainly required in order to produce reproducible polymers and make reproducible hydrogel contact lenses (both daily wear and continuous wear).

VI.2. CONCLUSIONS

Several conclusions can be made from the results obtained in this work.

(1) Poly(2-hydroxyethyl methacrylate) (i.e. PHEMA) hydrogel (equilibrium water content 40%) has "dissolved" oxygen permeability coefficients of 75×10^{-10} and 145×10^{-10} cc (S.T.P.).mm.cm^{-2}.sec^{-1}.cm Hg^{-1} at 25°C and 34°C respectively.

(2) Copolymerisations of HEMA with a hydrophobic monomer, such as styrene (i.e. St) or methyl methacrylate (i.e. MMA), at various compositions result in copolymers which become hydrogels on hydration if they contain more than about 75% HEMA in HEMA-St copolymers, or more than about 40% HEMA in HEMA-MMAcopolymers. These copolymer hydrogels have lower equilibrium water contents, lower "dissolved" oxygen permeability coefficients and smaller deformations under a load, than PHEMA hydrogel. Similar situations occur for HPA-St copolymers which become hydrogels on hydration if they contain more than about 60% HPA in the copolymers.

(3) Copolymerisations of HEMA or HPMA (i.e. hydroxypropyl methacrylate) with a more hydrophilic monomer (such as VP, i.e. N-vinyl pyrrolidone) at various compositions result in copolymers which are hydrogels on hydration. These copolymer hydrogels have higher equilibrium water contents, higher "dissolved" oxygen permeability coefficients and larger deformations under a load, than PHEMA hydrogel (or PHPMA hydrogel).

(4) Terpolymers of two hydrophilic components and one hydrophobic component (such as HEMA-VP-St and HPA-VP-St terpolymers) studied in this work are hydrogels on hydration, and their equilibrium water contents and "dissolved" oxygen permeability coefficients vary with the terpolymer compositions.

(5) Besides chemical structures and copolymer (or terpolymer) compositions other factors such as cross-link densities, hydration temperatures, polymerisation conditions and concentrations of sodium chloride solutions (hydrating solutions) also affect the equilibrium water contents of the hydrogels.

(6) Oxygen diffuses through the water in a hydrogel, and the "dissolved" oxygen permeability coefficients of the hydrogels increase exponentially with their equilibrium water contents. The empirical equations for correlating the "dissolved" oxygen permeability coefficients (P_d) and the equilibrium water contents (w) at 25°C and 34°C are respectively $P_d = 14 \times 10^{-10} \exp(0.0418w)$ and $P_d = 24 \times 10^{-10} \exp(0.0443w)$. Chemical structures(e.g. flexibility of the hydrophobic component) of the copolymers, however, also affect the "dissolved" oxygen permeability coefficients, particularly in the low-water-content hydrogels (i.e. hydrogels having equilibrium water contents $\lesssim$40%).

(7) The density and oxygen permeability results suggest three possible states of water in the hydrogels depending on the degree of hydration. In HEMA-St copolymers having equilibrium water contents less than about 10%, water is in the state of "primary" water (i.e. hydrogen-bonded to the polar group of the copolymers). As the degree of hydration is increased, "secondary" water starts to exist (i.e. water molecules hydrogen-bonded to the "primary" water). For those hydrogels having equilibrium water contents more than about 40% to 50%, "liquid-like" water seems to be present as well. In other words, the water exchange rate is higher in a hydrogel having a higher equilibrium water content.

(8) A higher-water-content hydrogel gives a larger deformation under a load than a lower-water-content hydrogel. Also, a hydrogel

containing flexible polymer chains is more deformable than a hydrogel containing rigid polymer chains. For high-water-content hydrogels, a small amount of a cross-linking agent (perhaps 1%) should be added so that both the deformation and recovery characteristics of the hydrogels are more elastic.

(9) A hydrogel made of two or more components is transparent provided (a) thay have similar reactivity ratios so that the dehydrated polymer is not heterogeneous, and (b) water is molecularly dispersed in the hydrogel and not just present as localised water clusters.

(10) Based on the literature values of the critical oxygen tensions of 15 mm.Hg at the corneal surface and the critical oxygen flux of 3.5 $\mu l.cm^{-2}.hr^{-1}$ into the cornea, it is estimated that a 0.1 mm thick contact lens for continuous wear should have a minimum "dissolved" oxygen permeability coefficient of 230×10^{-10} cc(S.T.P.). $mm.cm^{-2}.sec^{-1}.cm\ Hg^{-1}$ at 34°C, i.e. an equilibrium water content of at least 52% in 0.9% saline solution at 34°C. Of all the hydrogels studied in this work, hydrogels of HEMA-VP (50:50), HEMA-VP-St (47.5:47.5:5) and HPA-VP-St (35:55:10) satisfy the above requirements.

(11) A high-water-content hydrogel (equilibrium water content > 40%) has a density approaching that of water, a low refractive index approaching that of the tear fluid, a high "dissolved" oxygen permeability coefficient, and is expected to be more biocompatible with the eye than a low-water-content hydrogel. But on theother hand, a high-water-content hydrogel is more deformable under a load, and is predicted to be a better place for bacteria growth. It seems that a compromise of these properties has to be takan in selecting a contact lens material, for continuous wear in particular.

SUGGESTIONS FOR FURTHER WORK

Arising from this work, several suggestions for further work are made as follows:

(1) The effect of the chemical structure of the hydrophobic component in a high-water-content copolymer or terpolymer hydrogel (i.e. having an equilibrium water content > 40%) on its "dissolved" oxygen permeability coefficient, deformation properties and surface properties (such as wetting) have not been expermentially studied yet.

(2) The effect of the cross-link density in a high-water-content hydrogel on its properties relevant to contact lens applications should be investigated further, as a comparison with the effect of the chemical structure of the hydrophobic component, so that an optimised high-water-content hydrogel with a high oxygen permeability, elastic deformation and recovery, good dimensional stability and sufficient surface wettability will be ultimately developed.

(3) How the pH of a hydrating solution affecting the equilibrium water contents of hydrogels (High-water-content hydrogels in particular) should be studied as this is of importance in contact lens applications.

(4) More experimental methods (e.g. N.M.R., D.S.C. techniques) should be employed to study the states of water in the hydrogels, in order to correlate with the results (i.e. "primary", "secondary" and "liquid-like" waters) obtained in this work. This will give a better understanding of permeabilities of hydrogels to solutes such as oxygen.

(5) Should improvement of the small permeation cell used in this work be necessary to give a more accurate result on the measurement of "dissolved" oxygen permeability coefficient, the volume of the oxygen-

receiving chamber of the cell should be increased while still maintaining the sensitivity of the oxygen electrode in the system to an acceptable level, or a more accurate way of measuring the volume should be made.

(6) On the determination of cross-link densities of hydrogels using the compressive deformation property, the I.C.I. Microindentation Apparatus should be calibrated with the standard reticulometer used in the rubber work.

(7) Clinical studied should be carried out on contact lenses made of the existing high-water-content hydrogel materials studied in this work, so that (a) the results on the requirement of at least 230×10^{-10} cc(STP)mm.cm^{-2}.sec^{-1}.cm Hg^{-1} in the "dissolved" oxygen permeability coefficient at 34°C (i.e. at least 52% in equilibrium water content in 0.9% saline at 34°C) for a 0.1 mm.thick hydrogel contact lens for continuous wear and on the correlation between the "dissolved" oxygen permeability coefficient and the contact lens thickness could be justified, and (b) the visual acuity during contact lens wear could be correlated with the deformation properties studied here.

A P P E N D I C E S

(including tables and photographs)

APPENDIX 1.

(A) Derivation of the Equation (II.1) Correlating Equilibrium Water Content of a Hydrogel with its Density

If W_G, d_G and V_G are the weight, density and volume of a polymer in the hydrated state, W_p, d_p and V_p are the weight, density and volume of the polymer in the dehydrated state, and W_W, d_W and V_W are the weight, density and volume of water in the hydrogel, then

$$W_G = d_G V_G$$

$$= d_G(V_P + V_W)$$

$$= d_G \left(\frac{W_p}{d_p} + \frac{W_W}{d_W}\right)$$

i.e. $$\frac{d_G}{d_W} \cdot \frac{W_W}{W_G} = 1 - \left(\frac{d_G}{d_p} \cdot \frac{W_p}{W_G}\right)$$

$$= 1 - \frac{d_G}{d_p} + \left(\frac{d_G}{d_p} \cdot \frac{W_W}{W_G}\right)$$

$$\therefore \quad \frac{W_W}{W_G}\left(\frac{d_G}{d_W} - \frac{d_G}{d_p}\right) = 1 - \frac{d_G}{d_p}$$

$$\therefore \quad \frac{W_W}{W_G} = \left[\frac{\left(1 - \frac{d_G}{d_p}\right)}{d_G\left(\frac{1}{d_W} - \frac{1}{d_p}\right)}\right] \times 100\% \qquad \text{(II.1)}$$

(B) Derivation of the Equation (II.2) Correlating Equilibrium Water Content of a Hydrogel with its Refractive Index

If n_G, n_P and n_W are the refractive indices of a polymer in the hydrated state, in the dehydrated state, and of water in the hydrogel respectively, then

$$\frac{n_G^2 - 1}{n_G^2 + 2} = q_P \frac{n_P^2 - 1}{n_P^2 + 2} + q_W \frac{n_W^2 - 1}{n_W^2 + 2}$$

where q_P and q_W are the volume fractions of the polymer and water respectively in the hydrogel.

$$\therefore \quad \frac{1}{q_W}\left[\frac{n_G^2 - 1}{n_G^2 + 2} - \frac{n_P^2 - 1}{n_P^2 + 2}\right] = \frac{n_W^2 - 1}{n_W^2 + 2} - \frac{n_P^2 - 1}{n_P^2 + 2}$$

since $q_W = V_W/V_G$

or $\frac{1}{q_W} = 1 + \left[\frac{W_P}{d_P}\right]\left[\frac{1}{W_W}\right]$ (the density of water, d_W, at the room temperature was assumed to be 1)

then $\left[\frac{W_P}{W_W}\right]\left[\frac{n_G^2 - 1}{n_G^2 + 2} - \frac{n_P^2 - 1}{n_P^2 + 2}\right] = d_P\left[\frac{n_W^2 - 1}{n_W^2 + 2} - \frac{n_G^2 - 1}{n_G^2 + 2}\right]$

or $\left[\frac{W_P + W_W}{W_W}\right]\left[\frac{n_G^2 - 1}{n_G^2 + 2} - \frac{n_P^2 - 1}{n_P^2 + 2}\right] = d_P\left[\frac{n_W^2 - 1}{n_W^2 + 2} - \frac{n_G^2 - 1}{n_G^2 + 2}\right] +$

$$\left[\frac{n_G^2 - 1}{n_G^2 + 2} - \frac{n_P^2 - 1}{n_P^2 + 2}\right]$$

$$\therefore \quad \frac{W_W}{W_P+W_W} = \frac{W_W}{W_G} = \frac{\dfrac{n_G^2 - 1}{n_G^2 + 2} - \dfrac{n_P^2 - 1}{n_P^2 + 2}}{d_P\left[\dfrac{n_W^2 - 1}{n_W^2 + 2} - \dfrac{n_G^2 - 1}{n_G^2 + 2}\right] + \left[\dfrac{n_G^2 - 1}{n_G^2 + 2} - \dfrac{n_P^2 - 1}{n_P^2 + 2}\right]} \qquad \text{(II.2)}$$

APPENDIX 2 : TABLES OF CHAPTER II

TABLE II.1

Suppliers and purifications of the reagents used

Chemicals	Supplier	Purification
HEMA	B.D.H.	These reagents were purified in the conventional manner by reduced pressure distillations. The purity was checked by gas liquid chromatography
HPMA	Polysciences	
HEA	B.P.Chemicals	
HPA	Polysciences	
VP	Koch Light	
St	B.D.H.	
pMS	Koch Light	
MMA	B.D.H.	
4MPl	Koch Light	
EGDMA	B.D.H.	
BzP	Fisons Labs	Purified by recrystallisation
AZBN	B.D.H.	

TABLE II.2.

Preparation of hydroxyethyl methacrylate-styrene copolymer sheets

Sheet No.	Monomer feed (molar ratio) HEMA-St	Polymerisation conditions
1a.	100 : 0	BzP/72 hrs at 60°C/5 hrs at 100°C
1b	100 : 0	BzP (0.5%)/22 hrs at 65°C
1c	100 : 0	BzP/48 hrs at 65°C/6 hrs at 100°C
2	100 : 0 (+ 0.9% EGDMA)	BzP/72 hrs at 60°C/5 hrs at 100°C
3	95 : 5	BzP/70 hrs at 60°C/5 hrs at 100°C
4a	90 ; 10	BzP/70 hrs at 60°C/5hrs at 100°C
4b	90 : 10	AZBN (0.4%)/1½ hrs at 60°C
5	80 : 20	BzP/72 hrs at 60°C/4 hrs at 100°C
6	50 : 50	BzP/70 hrs at 60°C/6 hrs at 100°C
7	30 : 70	BzP/70 hrs at 60°C/6 hrs at 100°C
8	20 : 80	AZBN/65 hrs at 70°C
9	10 : 90	BzP/70 hrs at 60°C/6 hrs at 100°C
10	0 : 100	BzP/90 hrs at 60°C/6 hrs at 90°C

TABLE II.3

Preparation of hydroxypropyl methacrylate-styrene copolymer sheets

Sheet No.	Monomer feed (molar ratio) HPMA - St	Polymerisation conditions
11	100 : 0	BzP/24 hrs at 60°C
12	80 :20	AZBN/66 hrs at 70°C
13	50 : 50	BzP/32 hrs at 60°C/4 hrs at 90°C
14	20 : 80	AZBN/66 hrs at 70°C

TABLE II.4.

Preparation of hydroxyethyl acrylate-styrene copolymer sheets

Sheet No.	Monomer feed (molar ratio) HEA -St	Polymerisation conditions
15	70 : 30	BzP/72 hrs at 60°C/4 hrs at 100°C
16	65 : 35	BzP/48 hrs at 60°C/4 hrs at 100°C
17	60 : 40	BzP/75 hrs at 60°C
18	50 : 50	BzP/68 hrs at 60°C/6 hrs at 90°C
19	40 : 60	BzP/67 hrs at 60°C/6 hrs at 100°C
20	30 : 70	BzP/68 hrs at 60°C/6 hrs at 90°C
21	10 : 90	AZBN/72 hrs at 80°C

TABLE II.5.

Preparation of hydroxypropyl acrylate-styrene copolymer sheets

Sheet No.	Monomer feed (molar ratio) HPA - St	Polymerisation conditions
22	80 : 20	BzP/72 hrs at 60°C/6 hrs at 100°C
23	70 : 30	BzP/48 hrs at 60°C/4 hrs at 100°C
24	60 : 40	BzP/48 hrs at 60°C/4 hrs at 100°C
25	50 : 50	BzP/72 hrs at 60°C/6 hrs at 100°C
26	30 : 70	BzP/72 hrs at 60°C/6 hrs at 100°C
27	20 : 80	BzP/72 hrs at 60°C/6 hrs at 100°C
28	10 : 90	BzP/72 hrs at 60°C/6 hrs at 100°C

TABLE II.6.

Preparation of vinyl pyrrolidone-styrene copolymer sheets

Sheet No.	Monomer feed (molar ratio) VP - St	Polymerisation conditions
29	30 : 70	BzP/90 hrs at 60°C/6 hrs at 90°C
30	20 : 80	AZBN/72 hrs at 80°C
31	10 : 90	BzP/90 hrs at 60°C/6 hrs at 90°C
32	5 : 95	BzP/70 hrs at 60°C/6 hrs at 100°C

TABLE II.7.

Preparation of hydroxyethyl methacrylate-para-methyl styrene copolymer sheets

Sheet No.	Monomer feed (molar ratio) HEMA - pMS	Polymerisation conditions
33	90 : 10	BzP/70 hrs at 60°C/5 hrs at 100°C
34	10 : 90	AZBN/72 hrs at 70°C
35	0 : 100	AZBN/72 hrs at 70°C

TABLE II.8

Preparation of hydroxyethyl methacrylate-methyl methacrylate copolymer sheets.

Sheet No.	Monomer feed (molar ratio) HEMA - MMA	Polymerisation conditions
36	90 : 10	BzP/48 hrs at 65°C/6 hrs at 100°C
37	80 : 20	BzP/48 hrs at 65°C/6 hrs at 100°C
38	50 : 50	BzP/48 hrs at 65°C/6 hrs at 100°C

TABLE II.9.

Preparation of hydroxyethyl methacrylate-vinyl pyrrolidone copolymer sheets

Sheet No.	Monomer feed (molar ratio) HEMA -VP	Polymerisation conditions
39	90 : 10	BzP/45 hrs at 60°C/5 hrs at 100°C
40	70 : 30	BzP/45 hrs at 60°C/5 hrs at 100°C
41	50 : 50	BzP/45 hrs at 60°C/5 hrs at 100°C
42	40 : 60	BzP/45 hrs at 60°C/5 hrs at 100°C
43	20 : 80	BzP/45 hrs at 60°C/5 hrs at 100°C

TABLE II.10

Preparation of hydroxypropyl methacrylate-vinyl pyrrolidone copolymer sheets

Sheet No.	Monomer feed (molar ratio) HPMA - VP	Polymerisation conditions
44	0 : 100 (+ 5% EGDMA)	BzP/72 hrs at 60°C/17 hrs at 100°C
45	0 : 100 (+ 10% EGDMA)	BzP/72 hrs at 60°C/17 hrs at 100°C
46	50 : 50	AZBN/48 hrs at 60°C
47	50 : 50 (+ 0.6% EGDMA)	BzP/72 hrs at 70°C
48	50 : 50 (+ 1.2% EGDMA)	BzP/72 hrs at 70°C
49	80 : 20	AZBN/72 hrs at 65°C
50	80 : 20 (+ 1% EGDMA)	Bzp/72 hrs at 60°C/ 4 hrs at 90°C
51	80 : 20 (+ 5% EGDMA)	BzP/72 hrs at 60°C/4 hrs at 90°C
52	80 : 20 (+ 10% EGDMA)	BzP/72 hrs at 60°C/ 4 hrs at 90°C

TABLE II.11

Preparation of some terpolymer sheets

Sheet No.	Monomer feed (molar ratio)	Polymerisation conditions
53	HEMA-HPA-St (45:45:10)	BzP/48 hrs at 65°C/24 hrs at 100°C
54	HEMA-VP-St (45:45:10)	BzP/48 hrs at 65°C/24 hrs at 100°C
55	HPA-VP-St (45:45:10)	BzP/48 hrs at 65°C/24 hrs at 100°C

Note: 0.3% by wt. of the initiator (BzP or AZBN) was used in each case (Tables II.2 to II.11) unless indicated.

TABLE II.12.

Preparation of some copolymer rods

Rod.No.	Monomer feed (molar ratio)	Polymerisation conditions
R.1	HEMA-St (90:10)	0.5% BzP/8 hrs at 80°C/48 hrs at 90°C
R.2	HEMA-St (90: 10) + 0.5% malonic acid	0.5% BzP/4½ hrs at 80°
R.3	HEMA-St (90:10) + 1% malonic acid	0.5% BzP/4½ hrs at 80°
R.4	HPMA-VP (50:50)	0.4% BzP/9 hrs at 80°C/48 hrs at 100°
R.5	HPMA-VP (50:50) + 0.5% EGDMA	0.4% BzP/9 hrs at 80°C/48 hrs at 100°
R.6	HPA-VP (80:20) + 0.5% EGDMA	0.4% BzP/9 hrs at 80°
R.7	HPA-VP (80:20) + 1% EGDMA	0.4% BzP/4 hrs at 70°C/48 hrs at 100°

TABLE II.13

Solution copolymerisation of HEMA and 4MPl

Polymerisation mixture	Polymerisation condition	Method of obtaining polymer
(A) HEMA/4MPl/THF (25:25:50 by vol.)	65°C/1 hr	precipitated with MeOH
(B) HEMA/4MPl/THF (9:18:73 by vol.)	65°C/1 hr.	THF evaporated off.

TABLE II.14

Some properties of copolymer and terpolymer sheets when hydrated and dehydrated, (f = fusible; i = infusible; sol = soluble; insol = insoluble tp = transparent; tl = translucent; op = opaque; sl.tl = slightly translucent; r = rigid; sr = semi-rigid; fl = flexible; m.fl = moderately flexible; v.fl = very flexible; vw = very weak).

Sheet No	Material	When dehydrated				When hydrated		Successful preparation and isolation
		Fusion	Solubility	Transparency	Rigidity	Transparency	Rigidity	
1a,b, c.	PHEMA	i	insol	tp	r	tp	fl	yes
2	PHEMA (+0.9% EGDMA)	i	insol	tp	r	tp	fl	yes
3	HEMA-St (95:5)	i	insol	tp	r	tp	fl	yes
4a,b	HEMA-St (90:10)	i	insol	tp	r	tp	m.fl	yes
5	HEMA-St (80:20)	i	insol	tp	r	tp	sr	yes
6	HEMA-St (50:50)	i	insol	tp	r	tp	r	yes
7	HEMA-St (30:70)	i	insol	tp	r	tp	r	yes
8	HEMA-St (20:80)	i	insol	tp	r	tp	r	yes
9	HEMA-St (10:90)	f	sol	tp	r	tp	r	yes
10	Polystyrene	f	sol	tp	r	tp	r	yes
11	PHPMA	i	insol	tp	r	tp	m.fl	yes
12	HPMA-St (80:20)	-	-	-	-	-	-	No
13	HPMA-St (50:50)	i	insol	tp	r	tp	r	yes
14	HPMA-St (20:80)	i	insol	tp	r	tp	r	yes

TABLE II.14 (continued)

Sheet No.	Material	When dehydrated				when hydrated		Successful preparation and isolation
		Fusion	Solubility	Transparency	Rigidity	Transparency	Rigidity	
15	HEA-St (70:30)	i	insol	tp	sr	tp	fl	yes
16	HEA-St (65:35)	-	-	-	-	-	-	No
17	HEA-St (60:40)	i	insol	tp	sr	tp	sr	yes
18	HEA-St (50:50)	i	insol	tp	sr	tp	sr	yes
19	HEA-St (40:60)	i	insol	tp	r	tp	r	yes
20	HEA-St (30:70)	i	insol	tp	r	tp	r	yes
21	HEA-St (10:90)	f	sol	tp	r	tp	r	yes
22	HPA-St (80:20)	i	insol	tp	sr	tp	fl	yes
23	HPA-St (70:30)	i	insol	tp	sr	tp	fl	yes
24	HPA-St (60:40)	i	insol	tp	sr	tp	sr	yes
25	HPA-St (50:50)	i	insol	tp	sr	tp	sr	yes
26	HPA-St (30:70)	i	insol	tp	r	tp	r	yes
27	HPA-St (20:80)	i	insol	tp	r	tp	r	yes
28	HPA-St (10:90)	f	sol	tp	r	tp	r	yes
29	VP-St (30:70)	f	sol	tl	r	op	vw	No
30	VP-St (20:80)	f	sol	tp	r	op	vw	No
31	VP-St (10:90)	f	sol	tp	r	sl.tl	r	yes
32	VP-St (5:95)	f	sol	tp	r	tp	r	yes
33	HEMA-pMS (90:10)	i	insol	tp	r	tp	m.fl	yes
34	HEMA-pMS (10:90)	f	sol	sl.tl	r	sl.tl	r	yes
35	PpMS	f	sol	sl.tl	r	sl.tl	r	yes
36	HEMA-MMA (90:10)	i	insol	tp	r	tp	fl	yes
37	HEMA-MMA (80:20)	i	insol	tp	r	tp	m.fl	yes
38	HEMA-MMA (50:50)	i	insol	tp	r	tp	sr	yes

Table II.14 (continued)

Sheet No.	Material	When dehydrated: Fusion	Solubility	Transparency	Rigidity	When hydrated: Transparency	Rigidity	Suceessful preparation and isolation
39	HEMA-VP (90:10)	i	insol	tp	r	tp	fl	yes
40	HEMA-VP (70:30)	i	insol	tp	r	tp	fl	yes
41	HEMA-VP (50:50)	i	insol	tp	r	tp	v.fl	yes
42	HEMA-VP (40:60)	i	insol	tp	r	tp	v.fl	yes
43	HEMA-VP (20:80)	i	insol	tp	r	op	vw	yes
44	PVP (5% EGDMA)	i	insol	tp	r	sl.tl	w	yes
45	PVP (10% EGDMA)	i	insol	tp	r	sl.tl	w	yes
46	HPMA-VP (50:50)	i	insol	tp	r	sl.tl	v.fl	yes
47	HPMA-VP (50:50) + 0.6% EGDMA	i	insol	tp	r	tp	v.fl	yes
48	HPMA-VP (50:50) + 1.2% EGDMA	i	insol	tp	r	tp	v.fl	yes
49	HPMA-VP (80:20)	i	insol	tp	r	tp	fl	yes
50	HPMA-VP (80:20) + 1% EGDMA	i	insol	tp	r	tp	fl	yes
51	HPMA-VP (80:20) + 5% EGDMA	i	insol	tp	r	tp	m.fl	yes
52	HPMA-VP (80:20) + 10% EGDMA	i	insol	tp	r	tp	m.fl	yes
53	HEMA-HPA-St (45:45:10)	i	insol	tp	sr	tp	fl	yes
54	HEMA-VP-St (45:45:10)	i	insol	tp	r	tp	fl	yes
55	HPA-VP-St (45:45:10)	i	insol	tp	r	tp	fl	yes
56	HEMA-VP-St (47.5:47.5:5)	i	insol	tp	r	tp	fl	yes

Note: Sheets 36 to 62 were made by Dr.R.Enger.

TABLE II.14 (continued)

Sheet No.	Material	When dehydrated				When hydrated		Successful preparation and isolation
		Fusion	Solubility	Transparency	Rigidity	Transparency	Rigidity	
57	HPA-VP-St (35:35:30)	i	insol	tp	r	op	sr	yes
58	HPA-VP-St (75:15:10)	i	insol	tp	sr	tp	fl	yes
59	HPA-VP-St (55:35:10)	i	insol	tp	sr	tp	fl	yes
60	HPA-VP-St (35:55:10)	i	insol	tp	r	tp	v.fl	yes
61	HPA-VP-St (15:75:10)	i	insol	tp	r	tp	v.fl	yes
62	HPA-VP-St (47.5:47.5:5)	i	insol	tp	r	tp	v.fl	yes

Note: Sheets 56 to 62 were made by Dr.R.Ensor.

TABLE II.15

Properties of some copolymer rods

Rod No.	Material	Shrinkage during polymerisation	Mould release problem	Transparency	
				When dehydrated	When hydrated
R.1	HEMA-St (90:10)	15%	no	tp	tp
R.2	HEMA-St (90:10) + 0.5% malonic acid	12.5%	no	tp	tp
R.3	HEMA-St (90:10) + 1% malonic acid	6.4%	no	tp	tp
R.4	HPMA-VP (50:50)	6.7%	no	tp	sl.tl
R.5	HPMA-VP (50:50) + 0.5% EGDMA	6.2%	no	tp	tp
R.6	HPA-VP (80:20) + 0.5% EGDMA	-	severe	tp	tp
R.7	HPA-VP (80:20) + 1% EGDMA	-	severe	tp	tp

Note: tp = transparent; sl.tl = slightly translucent.

TABLE II 16.

Compression moulding trials of HEMA-St (90:10) and VP-HPMA (50:50) copolymer powders

Materials	Moulding temp. and time	Mould release aid	Size	Transparency	
				When dehydrated	When Hydrated
HEMA-St (90:10)	140°C/15 mins.	no	pieces	tp	op (0.9% saline
HEMA-St (90:10)	140°C/10 mins.	PVA	full	tp	op (2% saline)
HEMA-St (90:10)	190°C/15 mins	PVA	full	tp	tl (2% saline)
HEMA-St (90:10) + 10% succinic anhydride	160°C/60 mins.	no	pieces	tp	tp (water)
HEMA-St (90:10) + 20% succinic anhydride	160°C/60 Mins.	PVA	full	tp	tl (water)
VP-HPMA (50:50) + 1.6% succinic anhydride	140°C/20 mins.	PVA	full	tp	op (0.9% saline
VP-HPMA (50:50) + 10% malonic acid	150°C/3 mins	no	pieces	tp	op (0.9% saline

Note: tp = transparent; tl = translucent; op = opaque

TABLE II.17

Densities of some dehydrated and hydrated copolymers at 22°C

Sheet No.	Material	Density ($g.cc^{-1}$)	
		Dehydrated (d_p)	Hydrated (d_G)
1a	PHEMA	1.279	1.169
3	HEMA-St (95:5)	1.270	1.188
4a	HEMA-St (90:10)	1.259	1.204
5	HEMA-St (80:20)	1.242	1.215
6	HEMA-St (50:50)	1.174	1.176
7	HEMA-St (30:70)	1.130	1.131
9	HEMA-St (10:90)	1.090	1.090
10	Polystyrene	1.070	1.070
22	HPA-St (80:20)	1.195	1.137
23	HPA-St (70:30)	1.186	1.147
25	HPA-St (50:50)	1.146	1.147
26	HPA-St (30:70)	1.117	1.117
28	HPA-St (10:90)	1.082	1.086
36	HEMA-MMA (90;10)	1.263	1.185
37	HEMA-MMA (80:20)	1.249	1.200
38	HEMA-MMA (50:50)	1.233	1.205
39	HEMA-VP (90:10)	1.276	1.149
40	HEMA-VP (70:30)	1.269	1.120
41	HEMA-VP (50:50)	1.264	1.078
42	HEMA-VP (40:60)	1.266	1.047
44	PVP (5% EGDMA)	1.242	-
45	PVP (10% EGDMA)	1.245	-

TABLE II.18

Equilibrium water contents of some "hydrogels" at 22°C calculated from their densities, using Equation II.1 (d_W = 0.998)

Material	Density ($g.cc^{-1}$)		E.W.C. (W_W/W_G) (%)	
	d_p	d_G	Calculated (Eq.II.1)	Experimental (Chapter III)
PHEMA	1.279	1.169	33.4	36.0
HEMA-St (95:5)	1.270	1.188	25.3	29.7
HEMA-St (90:10)	1.259	1.204	17.6	22.8
HEMA-St (80:20)	1.242	1.215	9.1	13.1
HEMA-St (50:50)	1.174	1.176	0	5.0
HEMA-St (30:70)	1.130	1.131	0	2.5
HEMA-St (10:90)	1.090	1.090	0	1.3
HPA-St (80:20)	1.195	1.137	25.8	28.5
HPA-St (70:30)	1.186	1.147	18.1	17.2
HPA-St (50:50)	1.146	1.147	0	5.9
HPA-St (30:70)	1.117	1.117	0	1.4
HPA-St (10:90)	1.082	1.086	0	0.6
HEMA-MMA (90:10)	1.263	1.185	24.8	30.9
HEMA-MMA (80:20)	1.249	1.200	16.2	25.1
HEMA-MMA (50:50)	1.233	1.205	9.9	12.3
HEMA-VP (90:10)	1.276	1.149	39.7	40.6
HEMA-VP (70:30)	1.269	1.120	49.0	49.9
HEMA-VP (50:50)	1.264	1.078	64.7	66.1
HEMA-VP (40:60)	1.266	1.047	77.9	73.1

TABLE II.19

Refractive indices of some copolymers in both dehydrated and hydrated states at 22°C and 34°C

Material	Refractive index at 22°C		Refractive index at
	Dehydrated (n_D)	Hydrated (n_G)	34°C. Hydrated (n_G)
PHEMA	1.512	1.435	1.431
HEMA-St (95:5)	1.516	1.451	1.448
HEMA-St (90:10)	1.520	1.469	1.468
HEMA-St (80:20)	1.528	1.500	1.494
HEMA-St (50:50)	1.551	1.542	1.530
HEMA-St (30:70)	1.566	1.563	-
HEMA-St (10:90)	1.580	1.579	-
Polystyrene	1.589	1.589	-
PHPMA	-	1.465	1.465
HPA-St (80:20)	-	1.434	1.463
HEMA-pMS (90:10)	-	1.469	1.469
HEMA-MMA (90:10)	-	1.448	1.446
HEMA-MMA (80:20)	-	1.458	1.457
HEMA-MMA (50:50)	-	1.479	1.482
HEMA-VP (90:10)	-	1.429	1.427
HEMA-VP (70:30)	-	1.416	1.418
HEMA-VP (50:50)	-	1.394	1.398
HEMA-VP (40:60)	-	1.381	1.388
HPMA-VP (50:50)	-	-	1.395
HPMA-VP (50:50) + 0.6% EGDMA	-	-	1.430
HPMA-VP (50:50) + 1.2% EGDMA	-	-	1.435
HPMA-VP (80:20)	-	-	1.457

TABLE II.19 (continued)

Material	Refractive index at 22°C		Refractive index at
	Dehydrated (n_p)	Hydrated (n_G)	34°C. Hydrated (n_G)
HPMA-VP (80:20) + 1% EGDMA	-	-	1.461
HPMA-VP (80:20) + 5% EGDMA	-	-	1.466
HPMA-VP (80:20) + 10% EGDMA	-	-	1.473

TABLE II.20

Equilibrium water contents of HEMA-St "hydrogels" at 22°C calculated from their refractive indices, using Eq.II.2. (n_W = 1.333)

Material	Refractive index		d_p	E.W.C. (%)	
	n_G	n_P	(g/cc)	calculated (EQ.II.2)	Experimental Chapter III
PHEMA	1.435	1.512	1.279	35.6	36.0
HEMA-St (95:5)	1.451	1.516	1.270	28.9	29.7
HEMA-St (90:10)	1.469	1.520	1.259	21.7	22.8
HEMA-St (80:20)	1.500	1.528	1.242	11.3	13.1
HEMA-St (50:50)	1.542	1.551	1.174	3.2	5.0
HEMA-St (30:70)	1.563	1.566	1.130	1.0	2.5
HEMA-St (10:90)	1.579	1.580	1.090	0.3	1.3

(continued)

APPENDIX 3. TABLES OF CHAPTER III

TABLE III.1

Sheet No.	Material	E.W.C. (%) at 22°C
1a	PHEMA	36.0
2	PHEMA (0.9% EGDMA)	31.5
3	HEMA-St (95:5)	29.7
4a	HEMA-St (90:10)	22.8
5	HEMA-St (80:20)	13.1
6	HEMA-St (50:50)	5.0
7	HEMA-St (30;70)	2.5
8	HEMA-St (20:80)	1.4
9	HEMA-St (10:90)	1.3
10	Polystyrene	~ 0
11	PHPMA	22.6
13	HPMA-St (50:50)	3.6
14	HPMA-St (20:80)	2.1
15	HEA-St (70:30)	24.5
17	HEA-St (60:40)	13.0
18	HEA-St (50:50)	4.6
19	HEA-St (40:60)	3.2
20	HEA-St (30:70)	2.1
21	HEA-St (10:90)	1.0

(continued)

TABLE III.1. (continued)

Sheet No.	Material	E.W.C. (%) at 22°C
22	HPA-St (80:20)	28.5
23	HPA-St (70:30)	17.2
24	HPA-St (60:40)	-
25	HPA-St (50:50)	5.9
26	HPA-St (30:70)	1.4
27	HPA-St (20:80)	1.3
28	HPA-St (70:30)	0.6
31	VP-St (10:90)	1.4
32	VP-St (5:95)	0.4
33	HEMA-pMS (90;10)	24.8
36	HEMA-MMA (90:10)	30.9
37	HEMA-MMA (80:20)	25.1
38	HEMA-MMA (50:50)	12.3
39	HEMA-VP (90;10)	40.6
40	HEMA-VP (70:30)	49.9
41	HEMA-VP (50:50)	66.1
42	HEMA-VP (40:60)	73.2
44	PVP (5% EGDMA)	83.1
45	PVP (10% EGDMA)	74.1
46	HPMA-VP (50:50)	69.4
47	HPMA-VP (50:50) + 0.6% EGDMA	53.4
48	HPMA-VP (50:50) + 1.2% EGDMA	46.4
49	HPMA-VP (80:20)	33.6
50	HPMA-VP (80:20) + 1% EGDMA	31.5
51	HPMA-VP (80:20) + 5% EGDMA	25.4
52	HPMA-VP (80:20) + 10% EGDMA	20.6

(continued)

TABLE III.1. (continued)

Sheet No.	Material	E.W.C. (%) at 22°C
53	HEMA-HPA-St (45:45:10)	30.5
54	HEMA-VP-St (45:45:10)	42.7
55	HPA-VP-St (45:45:10)	61.6
56	HEMA-VP-St (47.5:47.5:5)	59.2
57	HPA-VP-St (35:35:30) (opaque)	13.5
58	HPA-VP-St (75:15:10)	45.6
59	HPA-VP-St (55:35:10)	55.5
60	HPA-VP-St (35:55:10)	78.9
61	HPA-VP-St (15:75:10)	78.4
62	HPA-VP-St (47.5:47.5:5)	74.4

TABLE III.2

Effect of sample drying conditions and sample thickness

Sheet No.	Material	E.W.C. (%) at 22°C	
		Sample dried at 65°C under vac. for 6 hrs.	Further drying at 65°c for 9 hours
1b	PHEMA (~0.2mm thick)	48.5	48.4
$1c_1$	PHEMA (~0.2mm thick)	43.0	42.7
$1c_2$	PHEMA (~0.4mm thick)	39.3	39.7
$1c_3$	PHEMA (~0.7mm thick)	36.2	36.6

TABLE III.3.

Effect of Hydration Temperature (* translucent; ‡ opaque)

Sheet No.	Material	E.W.C. (%) 22°C	34°C	40°C	50°C	60°C	70°C
1_a	PHEMA	36.0	37.4	37.3	42.0	40.8	43.0
2	PHEMA (0.9% EGDMA)	31.5	34.7	35.0	40.7	38.6	40.9
3	HEMA-St (95:5)	29.7	32.5	32.1	34.3	33.9	34.1
4 a	HEMA-St (90:10)	22.8	24.2	25.1	25.5	25.3	26.3
5	HEMA-St (80:20)	13.1	14.3	17.5	15.8	15.2	14.2
6	HEMA-St (50:50)	5.0	7.0	4.5	4.2	4.5	4.5
7	HEMA-St (30:70)	2.5	2.7	2.3	2.9	2.3	1.6
9	HEMA-St (10:90)	1.4	~0	0.7	~0	~0	~0
11	PHPMA	22.6	20.9	20.6	25.1	40.8*	78.7*
22	HPA-St (80:20)	28.5	24.8	24.1	40.3	48.1	-
23.	HPA-St (70:30)	17.2	13.0	13.0	21.6	-	55.6*
25	HPA-St (50:50)	5.9	5.1	4.6	10.1‡	22.8‡	29.6‡
26	HPA-St (30:70)	1.4	2.2	1.4	1.4	1.1	1.4
28	HPA-St (10:90)	0.6	0.9	0.3	0.3	0.3	0.3
36	HEMA-MMA (90:10)	30.9	31.9	32.3	39.9	50.6	56.3
37	HEMA-MMA (80:20)	25.1	24.1	24.1	31.1	37.1	41.8
38	HEMA-MMA (50:50)	12.3	11.3	12.5	11.5	10.7	10.6
39.	HEMA-VP (90:10)	40.6	41.4	41.6	45.2	45.0	45.6
40	HEMA-VP (70:30)	49.9	46.6	46.1	46.8	44.7	44.4
41.	HEMA-VP (50:50)	66.1	56.4	56.6	55.5	50.0	52.5
42	HEMA-VP (40:60)	73.2	64.4	63.2	59.9	56.0	64.5
44	PVP (5% EGDMA)	83.1	81.2	81.8	83.0	-	-
45	PVP (10% EGDMA)	74.1	73.8	73.2	72.6	73.5	75.0
46	HPMA-VP (50:50)	69.4	59.6	60.7	71.3	86.1	87.0
47	HPMA-VP (50:50) + 0.6% EGDMA	53.4	46.6	48.5	-	61.1	-
48	HPMA-VP (50:50) + 1.2% EDGMA	46.4	41.4	43.2	60.4	61.0	-

(continued)

TABLE III.3 (continued)

Sheet No.	Material	E.W.C. (%) 22°C	34°C	40°C	50°C	60°C	70°C
49	HPMA-VP (80:20)	33.6	28.2	27.8	26.2	35.8*	78.1*
50	HPMA-VP (80:20) + 1% EGDMA	31.5	25.4	24.7	23.5	35.6	39.0
51	HPMA-VP (80:20) + 5% EGDMA	25.4	23.6	22.0	24.1	-	27.6
52	HPMA-VP (80:20) + 10% EGDMA	20.6	23.3	20.2	23.6	26.0	-

TABLE III.4.

Effect of concentration of NaCl solution

Sheet No.	Material	E.W.C. (%) at 22°C Distilled water	0.9% saline	1.5% saline	2.0% saline
$1c_2$	PHEMA	39.9	39.0	37.8	36.8
4b	HEMA-St (90:10)	26.0	24.5	23.0	22.5
55	HPA-VP-St (45:45:10)	55.7	52.0	49.4	49.7

The water content (W.C.) and fractional weight of equilibrium water $(W_W)_t/W_W)_\infty$ during hydration and dehydration of each of PHEMA, HEMA-St (90:10) and HEMA-VP (50:50) copolymers are shown in Tables III.5 and III.6 respectively.

TABLE III.5.

Hydration studies of some polymers

Time of hydration (mins)	PHEMA (0.15 mm thick)		HEMA-St (90:10) *		HEMA-VP (50:50) ‡	
	W.C. (%)	$(W_W)_t/(W_W)_\infty$	W.C.(%)	$(W_W)_t/(W_W)_\infty$	W.C.(%)	$(W_W)_t/(W_W)_\infty$
0	0	0	0	0	0	0
2	13.4	0.35	4.3	0.17	32.4	0.49
5	21.0	0.55	-	-	44.0	0.66
10	31.9	0.84	11.9	0.48	55.7	0.84
20	38.0	1.0	17.7	0.71	63.6	0.96
30	38.0	1.0	22.4	0.91	65.7	0.99
40	37.1	0.98	23.8	0.96	66.6	1.0
50	38.0	1.0	24.8	1.0	66.6	1.0
60	38.0	1.0	24.8	1.0	66.6	1.0

* 0.21 mm. thick

‡ 0.15 mm. thick

TABLE III.6.

Dehydration studies of some hydrogels

Time (mins) of dehydration	PHEMA		HEMA-St (90:10)		HEMA-VP (50:50)	
	W.C.(%)	$(W_W)_t/(W_W)_o$	W.C.(%)	$(W_W)_t/(W_W)_o$	W.C.(%)	$(W_W)_t/(W_W)_o$
0	38.7	1.0	25.0	1.0	65.8	1.0
3	34.2	0.86	21.9	0.88	60.7	0.92
5	31.6	0.82	20.6	0.83	58.0	0.88
10	23.4	0.61	17.4	0.70	52.3	0.80
15	17.4	0.45	15.1	0.61	46.3	0.70
20	12.6	0.33	13.3	0.54	39.8	0.60
25	-	-	12.0	0.48	35.3	0.54
30	8.4	0.22	10.5	0.42	31.6	0.48
35	7.8	0.20	9.4	0.38	-	-
40	6.3	0.16	8.8	0.35	22.6	0.34
45	5.4	0.14	8.4	0.34	19.4	0.29
50	5.1	0.13	8.2	0.33	16.1	0.25
55	4.8	0.12	7.5	0.30	11.8	0.18
60	4.8	0.12	6.7	0.27	9.2	0.14
65	4.8	0.12	6.2	0.25	7.4	0.11
70	4.5	0.12	6.0	0.24	5.5	0.08
75	4.2	0.11	6.2	0.25	4.6	0.07
80	4.2	0.11	5.8	0.23	4.3	0.066
85	4.8	0.12	5.8	0.23	3.8	0.058
90	4.8	0.12	5.6	0.22	3.6	0.055

TABLE III.7

Volume swell (V_G/V_P) and linear swell (L_G/L_P) of HEMA-St copolymers on hydration, using Equations III.2 and 3.

HEMA-St Monomer feed (mole %)	(V_G/V_P) (%) (Equation III.2)	(L_G/L_P) (%) (Equation III.3)
100 : 0	170.9	119.6
95 : 5	152.2	115.0
90 : 10	135.4	110.7
80 : 20	117.8	105.5
50 : 50	105.1	101.8
30 : 70	102.0	100.8
10 : 90	101.2	100.2

TABLE III.8

Volume swell (V_G/V_P) and linear swell (L_G/L_P) of HEMA-VP copolymer hydrogels, using Equations III.2 and 3

HEMA-VP monomer feed (mole %)	(V_G/V_P) (%) (Equation III.2)	(L_G/L_P) (%) (Equation III.3)
90 : 10	187.0	123.0
70 : 30	226.0	131.2
50 : 50	346.0	151.2
40 : 60	451.0	165.1

TABLE III.9

Results calculated from the plot of $\ln\left(1 - \frac{(W_W)_t}{(W_W)_\infty}\right)$

$$\ln\left(1 - \frac{(W_W)_t}{(W_W)_\infty}\right) = A - kt$$

Material	A	k (min^{-1})
PHEMA	ln 9.2	0.1524
HEMA-VP (50:50)	ln 6.8	0.1459
HEMA-St (90:10)	ln 9.4	0.0602

APPENDIX 4

TABLES OF CHAPTER IV.

TABLE IV.1

Material	T(°C)	P_g (cc(STP).mm.cm^{-2}.sec^{-1}.cm Hg^{-1})
poly(4-methyl pent-1-ene)	20	183 x 10^{-10}
	25	226 x 10^{-10}
	34	282 x 10^{-10}
polystyrene	20	29.6 x 10^{-10}
	25	36.8 x 10^{-10}

TABLE IV. 2

"GO" permeability coefficients of some hydrophobic polymer sheets at 22°C under the oxygen-to-air conditions, using the modified E.I.L. electrode system.

Material	P_g (cc(STP).mm.cm^{-2}.sec^{-1}.cm Hg^{-1}) at 22°C		
	Commercial rubber "O" ring	Metal washer	"Home-made" flat rubber "O" ring
Poly(4-methyl pent-1-ene) *	115×10^{-10}	90×10^{-10}	120×10^{-10}
Oriented polypropylene *	-	15.7×10^{-10}	5.7×10^{-10}
Silicone rubber ‡	1050×10^{-10}	-	-
E.P.T. **	-	-	126×10^{-10}

* obtained from the Metal Box Ltd.

‡ in a lens form, from the Department of Ophthalmic Optics

** cross-linked with 1% dicumyl peroxide at 160°C for 5 mins.

TABLE IV 3.

Reported "GO" permeability coefficients of some hydrophobic polymer sheets at 25°C

Material	P_g(cc(STP).mm.cm^{-2}.sec^{-1}.cm Hg^{-1})	Reference
poly(4-methyl pent-1-ene)	275 x 10^{-10}	213
Oriented polypropylene	7 x 10^{-10}	213
Silicone rubber	5000 x 10^{-10}	131

TABLE IV.4

Calibrations of the I.M.I. oxygen sensor in gas and in water

Gas	Monitor reading		
	Calibration in gas	Calibration in gas-saturated water	
		Stirred	Unstirred
nitrogen	0	0	0
air	21	16.5	8
oxygen	100	80	39

TABLE IV.5.

"GO" permeability coefficients of poly(4-methyl pent-1-ene) and ethylene propylene terpolymer at 22°C under air-to-nitrogen condition using the I.M.I. oxygen sensor and the large permeation cell.

Material	P_g(cc(STP).mm.cm^{-2}.sec^{-1}.cm Hg^{-1}) at 22°
poly(4-methyl pent-1-ene)	274 x 10^{-10}
ethylene propylene terpolymer	183 x 10^{-10}

TABLE IV.6

"GO" permeability coefficients of poly(4-methyl pent-1-ene) and HEMA-St (90:10) hydrogel at 34°C under water/water conditions, using the I.M.I. oxygen sensor and the large permeation cell

Material	L (mm)	P_d(cc(STP).mm.cm^{-2}.sec^{-1}.cm Hg^{-1}) at 34°C
poly(4-methyl pent-1-ene)	0.25	198 x 10^{-10}
HEMA-St (90:10)	0.15	55 x 10^{-10}

TABLE IV.7

"GO" permeability coefficients of some hydrophobic materials under air-to-nitrogen conditions at 22°C, using the I.M.I. oxygen sensor and the small permeation cell

Material	L (mm)	P_g(cc(STP).mm.cm^{-2}.sec^{-1}.cm Hg^{-1}) at 22°C
poly(4-methyl pent-1-ene)	0.23	215 x 10^{-10}
polystyrene	0.19	37 x 10^{-10}

TABLE IV.8

"DO" permeability coefficients of some hydrophobic materials at various thicknesses and at 25°C and 34°C (under water/water conditions.)

Material	L (mm)	P_d(cc(STP)mm.cm^{-2}.sec^{-1}.cm Hg^{-1}) at 25°C	at 34°C
poly(4-methyl pent-1-ene)	0.075	116 x 10^{-10}	132 x 10^{-10}
	0.23	181 x 10^{-10}	226 x 10^{-10}
	0.40	192 x 10^{-10}	250 x 10^{-10}
polystyrene	0.16	40.6 x 10^{-10}	40.8 x 10^{-10}
	0.19	-	42.4 x 10^{-10}

TABLE IV. 9

$(1/P_d)$ and $(1/L)$ values for poly(4-methyl pent-1-ene) at 25°C and 34°C

L (cm)	$\frac{1}{L}$ (cm^{-1})	P_d ($cc(STP).mm.cm^{-2}.sec^{-1}.cm\ Hg^{-1}$)		$(1/P_d)$ ($cc(STP)^{-1}.mm^{-1}.cm^2.sec.cm\ Hg$)	
		25°C	34°C	25°C	34°C
$0.75X10^{-2}$	$1.333x10^{2}$	$116x10^{-10}$	$132x10^{-10}$	$0.862x10^{8}$	$0.757x10^{8}$
$2.3x10^{-2}$	$0.435x10^{2}$	$181x10^{-10}$	$226x10^{-10}$	$0.552x10^{8}$	$0.442x10^{8}$
$4.0x10^{-2}$	$0.250x10^{2}$	$192x10^{-10}$	$250x10^{-10}$	$0.521x10^{8}$	$0.400x10^{8}$

TABLE IV.10

(DS) values obtained from Figurs IV.10 for poly(4-methyl pent-1-ene) at 25°C and 34°C, and their comparisons with P_g values

Temperature (°C)	(D.S) ($cc(STP).mm.cm^{-2}\ sec^{-1}.cm\ Hg^{-1}$) (from Figure IV.10)	P_g ($cc(STP).mm.cm^{-2}\ sec^{-1}.cm\ Hg^{-1}$) (from Davenport Gas Permeability Apparatus)	P_g ($cc(STP).mm\ sec^{-1}.cm\ Hg^{-1}$) (from the Metal Box Limited)
25	250×10^{-10}	226×10^{-10}	275×10^{-10}
34	322×10^{-10}	282×10^{-10}	-

TABLE IV.11

Effect of surface treatment of poly(4-methyl pent-1-ene) on its "dissolved" oxygen permeability coefficient at 34°C

Poly(4-methyl pent-1-ene	L (mm)	P_d ($cc(STP).mm.cm^{-2}.sec^{-1}.cm\ Hg^{-1}$) at 34°C
Untreated	0.23	238×10^{-10}
	0.23	229×10^{-10}
Treated *	0.23	256×10^{-10}

* treated with chromic acid at 60°C for 15 mins.

TABLE IV.12

"DO" permeability coefficients of some hydrogels at 34°C measured by the "laminate" method of a poly(4-methyl pent-1-ene) sheet and the hydrogel sample.

Hydrogel	With a TPX* sheet support			With no TPX sheet support	
	L_{Gel} (mm)	L_{TPX} (mm)	$(P_d)_{Gel}$ ‡ units	L_{Gel} (mm)	$(P_d)_{Gel}$ units
PHEMA	0.21	0.075	190×10^{-10}	0.23	179×10^{-10}
	0.23	0.075	182×10^{-10}		
	0.21	0.23	146×10^{-10}		
	0.21	0.40	146×10^{-10}		
HEMA-St(90:10)	0.21	0.075	70×10^{-10}	0.19	72×10^{-10}
	0.18	0.23	62×10^{-10}		
HEMA-St(80:20)	0.11	0.23	21×10^{-10}	-	-
HEMA-pMS(90:10)	0.21	0.23	49×10^{-10}	0.21	56×10^{-10}
HPA-St(80:20)	0.17	0.23	77×10^{-10}	0.18	70×10^{-10}
HEMA-MMA(90:10)	0.21	0.23	73×10^{-10}	-	-
HEMA-MMA(80:20)	0.17	0.23	54×10^{-10}	-	-
HEMA-VP(70:30)	0.19	0.23	125×10^{-10}	-	-
HEMA-VP(50:50)	0.22	0.075	264×10^{-10}	0.22	284×10^{-10}
	0.22	0.23	173×10^{-10}		
HPA-VP-St (45:45:10)	0.27	0.075	238×10^{-10}	0.24	259×10^{-10}
HPA-VP-St (35:55:10)	0.26	0.075	341×10^{-10}	0.24	393×10^{-10}
	0.26	0.40	233×10^{-10}		

* TPX is the trade name for poly(4-methyl pent-1-ene).

‡ In this case, $(P_d)_{Gel}$ was calculated from k_{Gel} where k_{Gel} was assumed to be equal to $k_{Laminate}$.

"units" here means $cc(STP).mm.cm^{-2}.sec^{-1}.cm\ Hg^{-1}$.

TABLE IV.13

Effect of water flow rate during measurement on the "DO" permeability coefficient of PHEMA hydrogel at 25°C

Water flow rate (ml/sec)	P_d (cc(STP).mm.cm^{-2}.sec^{-1}.cm Hg^{-1})	E.W.C. (%)
~ 2	98 x 10^{-10}	48.5
~ 8	95 x 10^{-10}	48.5

TABLE IV.14

Effect of hydrogel sample thickness on the "DO" permeability coefficient of PHEMA hydrogel at 25°C and 34°C

Sample thickness L (mm)	P_d (cc(STP).mm.cm^{-2}.sec^{-1}.cm.Hg^{-1})	E.W.C. (%)	Temperature (°C)
0.21	77 x 10^{-10}	40.5	25
0.39	72 x 10^{-10}	39.5	25
0.21	145 x 10^{-10}	40.0	34
0.37	147 x 10^{-10}	40.0	34

TABLE IV.15

"DO" permeability coefficients of some hydrogels and non-hydrogels at 25°C, using the "direct" sample mounting method (i.e. no TPX sheet support)

Units of P_d : cc(STP).mm.cm^{-2}.sec^{-1}.cm Hg^{-1}

Material	L (mm)	P_d units	E.W.C. (%)
PHEMA	0.39	72 x 10^{-10}	39.5
	0.21	77 x 10^{-10}	40.5
	0.19	91 x 10^{-10}	42.8
	0.21	98 x 10^{-10}	48.5
HEMA-St (95:5)	0.16	49 x 10^{-10}	29.7
HEMA-St (90:10)	0.20	35 x 10^{-10}	22.8

(continued)

TABLE IV.15 (continued)

Material	L (mm)	P_d (units)	E.W.C. (%)
HEMA-St (80:20)	0.11	9.2×10^{-10}	13.1
HEMA-St (50:50)	0.10	2.5×10^{-10}	5.0
HEMA-St (10:90)	0.14	14.4×10^{-10}	1.4
polystyrene	0.16	40.6×10^{-10}	0
HEMA-MMA (90:10)	0.23	54.1×10^{-10}	30.9
HEMA-MMA (80:20)	0.13	42.2×10^{-10}	25.1
HEMA-MMA (50:50)	0.14	12.7×10^{-10}	12.3
HEMA-pMS (90:10)	0.21	37.5×10^{-10}	24.8
HEMA-EA (50:50) *	0.19	37.9×10^{-10}	15.7
HEMA-EHMA (50:50) **	0.22	14.0×10^{-10}	5.6
HPA-St (80:20)	0.16	73.8×10^{-10}	36.2
HEA-St (60:40)	0.24	20.2×10^{-10}	13.0
HEMA-acrylamide-methacrylic acid (25:25:50)	0.24	40.5×10^{-10}	35.3
PHPMA	0.15	17.7×10^{-10}	22.6
HEMA-VP (70:30)	0.23	128×10^{-10}	50.0
HEMA-VP (50:50)	0.19	230×10^{-10}	66.0
HEMA-VP (40:60)	0.23	261×10^{-10}	73.0
HEMA-HPA-St (45:45:10)	0.20	56×10^{-10}	30.5
HEMA-VP-St (45:45:10)	0.23	97×10^{-10}	42.7
HEMA-VP-St (47.5:47.5:5)	0.38	200×10^{-10}	59.2
HPA-VP-St (35:35:30)	0.21	6×10^{-10}	13.5
HPA-VP-St (75:15:10)	0.16	96×10^{-10}	45.6
HPA-VP-St (55:35:10)	0.24	131×10^{-10}	55.5
HPA-VP-St (45:45:10)	0.23	204×10^{-10}	61.6
HPA-VP-St (35:55:10)	0.26	373×10^{-10}	78.9
HPA-VP-St (15:75:10)	0.34	410×10^{-10}	78.4

(continued)

TABLE IV.15 (continued)

Material	L (mm)	P_d (units)	E.W.C. (%)
HPA-VP-St (47.5:47.5:5)	0.24	354×10^{-10}	74.4

* EA ; ethyl acrylate

** EHMA : 2-ethyl hexyl methacrylate

TABLE IV.16

"DO" permeability coefficients of some hydrogels and non-hydrogels at 34°C, using the "direct" sample mounting method (i.e. No TPX sheet support). Units of P_d : cc(STP).mm.cm^{-2}.sec^{-1}.cm Hg^{-1}

Material	L (mm)	P_d (units)	E.W.C. (%)
PHEMA	0.21	145×10^{-10}	40
	0.37	147×10^{-10}	40
PHEMA	0.23	179×10^{-10}	46.7
HEMA-St (90:10)	0.18	68×10^{-10}	24.2
	0.21	76×10^{-10}	24.2
HEMA-St (50:50)	0.10	6.3×10^{-10}	6.0
HEMA-St (30:70)	0.13	9.5×10^{-10}	2.7
HEMA-St (10:90)	0.14	9.4×10^{-10}	~0
Polystyrene	0.16	40.8×10^{-10}	~0
HEMA-pMS (80:20)	0.21	56×10^{-10}	21.2
HPA-St (80:20)	0.18	70×10^{-10}	24.8
HEMA-VP (50:50)	0.22	284×10^{-10}	56.4
HEMA-VP-St (47.5:47.5:5)	0.39	303×10^{-10}	51.2
HPA-VP-St (45:45:10)	0.24	259×10^{-10}	48.6
HPA-VP-St (35:55:10)	0.24	393×10^{-10}	64.5

APPENDIX 5. TABLES OF CHAPTER V.

TABLE V.1a.

The 1-minute deformations of some hard materials under various loads, a spherical-end indentor (1/16 inch diameter) being used. (h = depth of deformation; t = sample thickness)

Load m gm	1-minute deformation, h (microns)				
	PMMA * (t=0.18mm)	Polystyrene (t=0.15mm)	TPX‡ (t=0.22mm)	Dehydrated (t=0,15mm)	Dehydrated HEMA-St(90:10) (t=0.20mm)
1	0.15	0.14	0.23	0.15	0.13
2	0.25	0.27	0.35	0.25	0.25
4	0.37	0.47	0.58	0.45	0.40
8	0.63	0.75	0.94	0.65	0.65
16	1.08	1.15	1.48	1.00	0.95

* Poly(methyl methacrylate)

‡ Poly(4-methyl pent-1-ene)

TABLE V.1b

The 1-minute deformations of some soft materials under various loads, a spherical indentor (1/16 inch diameter) being used .

(h in micron)

Load m (gm)	EPT *		SBR‡ (t=0.45 mm)		Hydrated PHEMA (t=0.18 mm)		hydrated HEMA-St(90:10) (t=0.22 mm)	
	h (micron)	(h/(t-h))	h (micron)	(h/(t-h))	h	(h/(t-h))	h	(h/(t-h))
0.14	-	-	-	-	2.39	1.35×10^{-2}	1.67	0.76×10^{-2}
0.24	2.50	1.26×10^{-2}	3.64	0.82×10^{-2}	3.54	2.05×10^{-2}	3.05	1.40×10^{-2}
0.34	3.03	1.54×10^{-2}	4.85	1.09×10^{-2}	4.70	2.68×10^{-2}	4.03	1.86×10^{-2}
0.48	-	-	-	--	5.60	3.21×10^{-2}	4.71	2.19×10^{-2}
0.75	5.74	2.95×10^{-2}	-	-	7.12	4.12×10^{-2}	6.51	3.05×10^{-2}
1.0	6.62	3.42×10^{-2}	9.40	2.14×10^{-2}	-	-	-	-
2.0	10.0	5.26×10^{-2}	15.6	3.60×10^{-2}	-	-	-	-
4.0	15.2	8.23×10^{-2}	23.7	5.55×10^{-2}	-	-	-	-

* Ethylene propylene terpolymer

‡ Styrene butadiene rubber

TABLE V.3.

Deformation measurements of dimethyl sulphoxide -swollen PHEMA gel under various loads, using a flat-end indentor

Load F (gm)	1-minute deformation (microns)
0.34	1.50
0.75	2.70
1.0	3.10
2.0	4.65
4.0	7.30

A P P E N D I X 6

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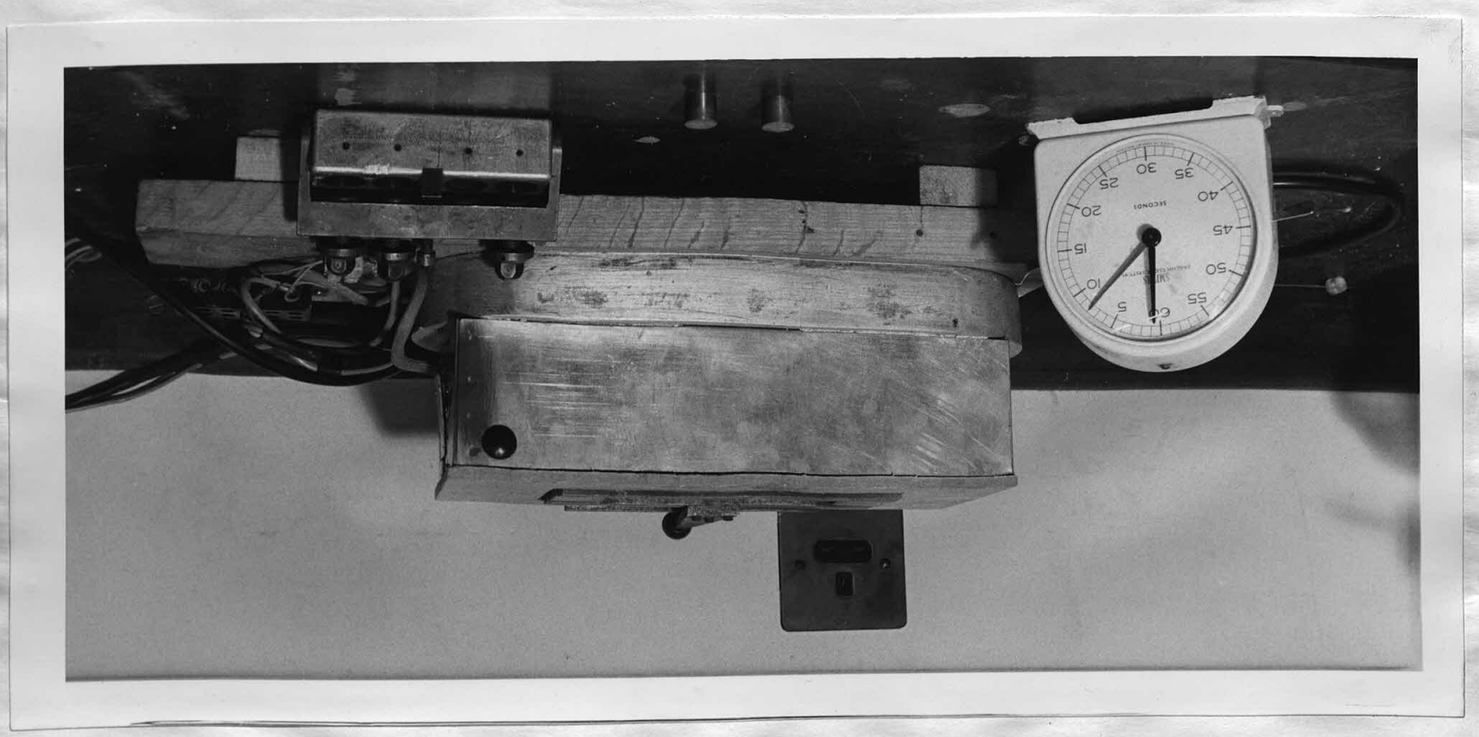


Plate II.1 The modified Kelvin's compression moulding apparatus.

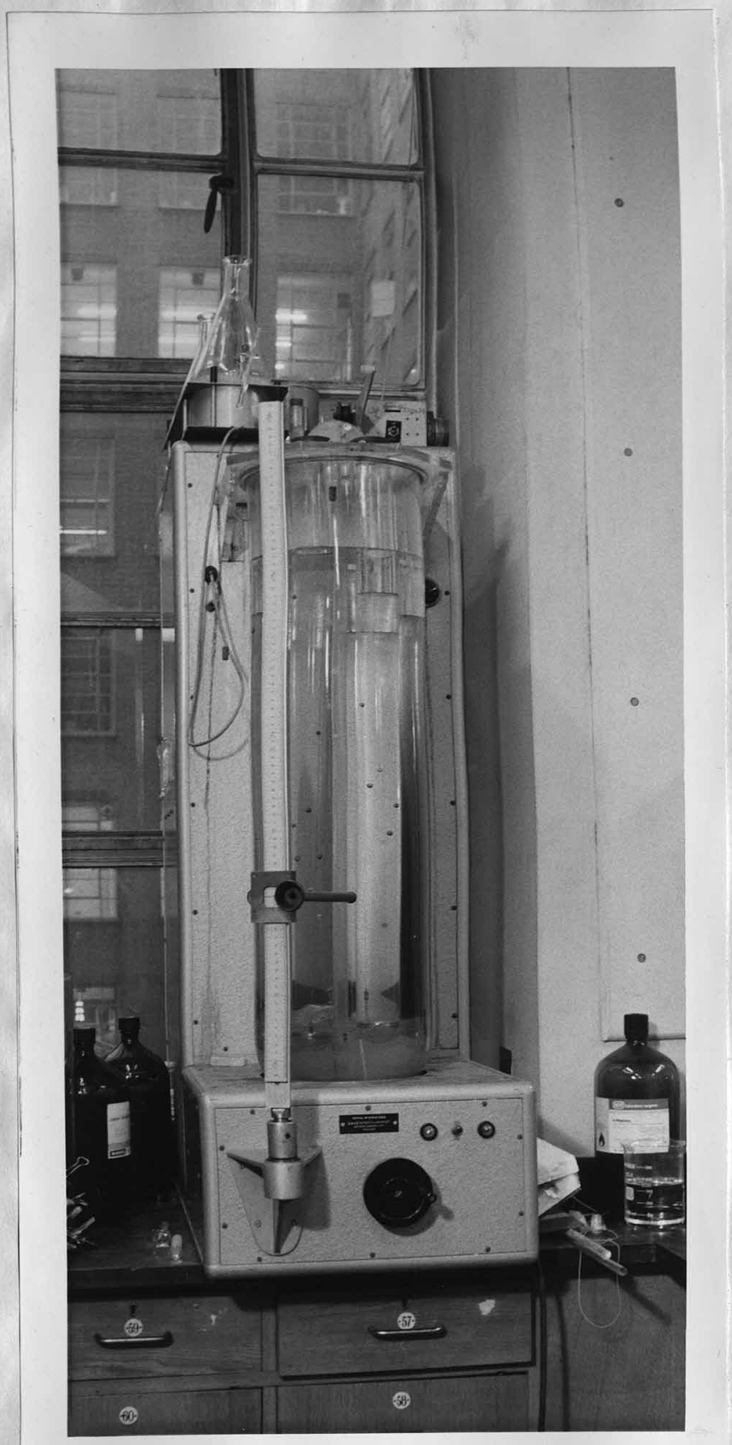


Plate II.2 The Davenport density gradient columns.



Plate II.3 The Abbe refractometer.

Plate IV.1 The Davenport gas permeability apparatus

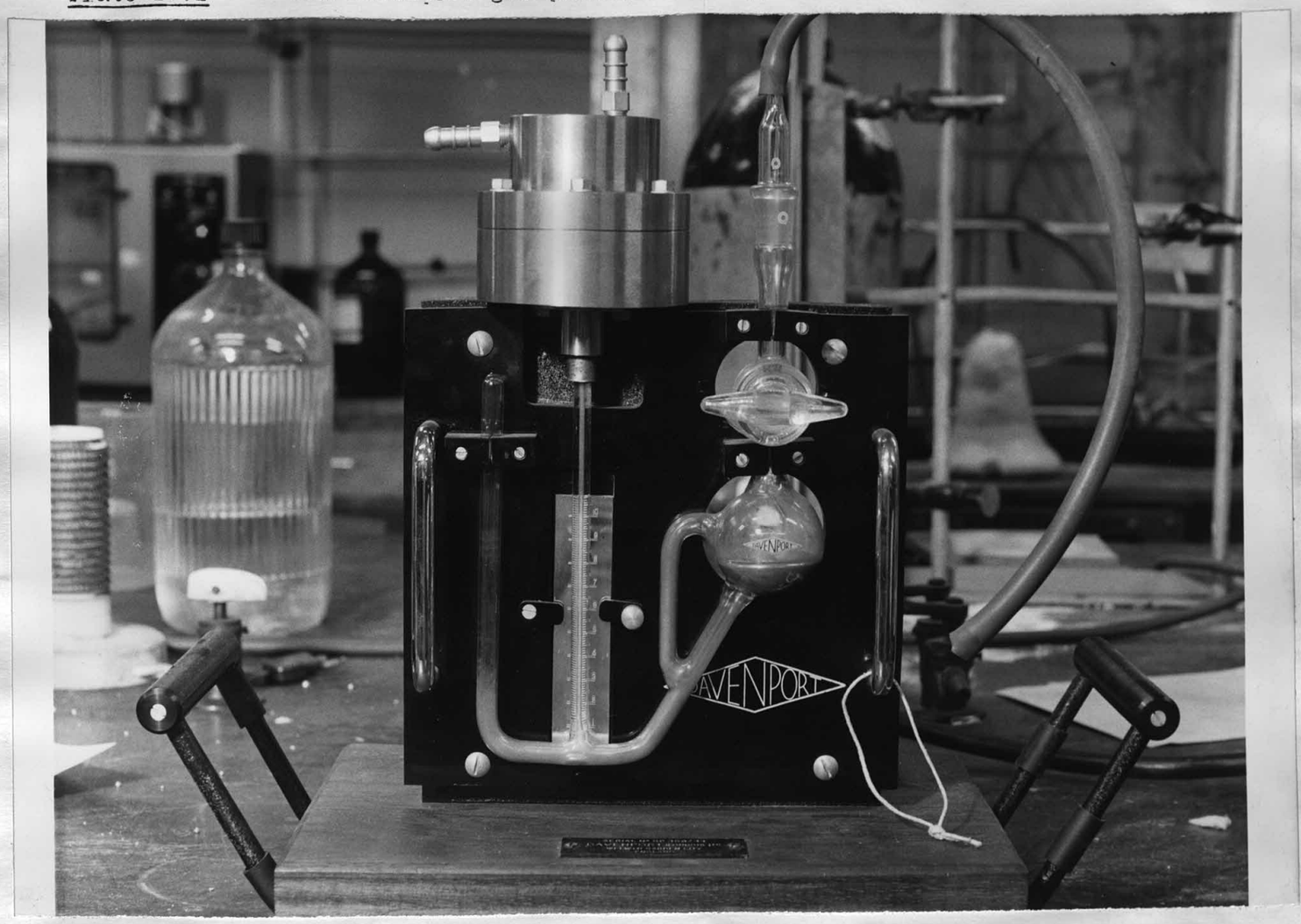
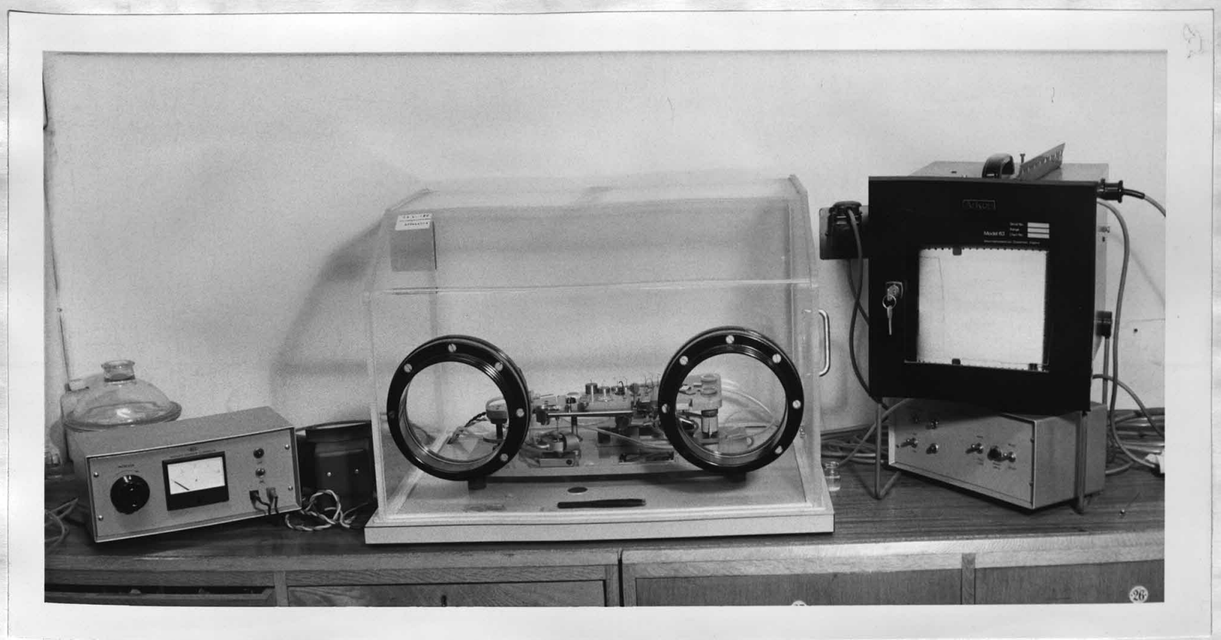


Plate IV.2 The E.I.L. polarographic oxygen sensor.

Reference Target LRT-8E4-V3

Plate IV.4 The layout of the "dissolved" oxygen permeability measuring apparatus.

Plate V.1 The layout of the I.C.I. Micro-indentation Hardness Apparatus.



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