STEREOPHOTOGRAMMETRIC MAPPING OF THE ANTERIOR SURFACE OF THE HUMAN CORNEA

Dr Jon Osborn, Department of Surveying and Spatial Information Science, University of Tasmania, Australia
Dr Gordon Wise, Senior Ophthalmologist, Tasmanian Eye Clinics, Wentworth St, South Hobart, Tasmania, Australia

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ABSTRACT

The major refractive element of the human eye is the anterior surface of the cornea. An accurate knowledge of the topography of this surface is important in a variety of ophthalmic applications. Existing commercially available methods of mapping corneal topography rely on measuring the images of mires reflected from the corneal surface. These methods are inadequate, particularly for abnormal corneas and intra-surgical application.
This paper describes a stereophotogrammetric method for measuring corneal topography. Photogrammetric verification of the method shows that accuracy in the order of 20μm (r.m.s) has been achieved. The instrument has been installed at an eye diagnostic clinic and tested on patients with corneal abnormalities. Some results are illustrated and contrasted with corresponding photokeratoscope images.
The design of a digital implementation of the instrument is described, including details such as camera type, array size and resolution, target recognition and camera calibration. The clinical parameters that affect the commercial viability of a digital system are discussed.

1. CORNEAL TOPOGRAPHY

The topography of the anterior surface of the human cornea is complex. Descriptions usually divide it into zones — the central, paracentral, peripheral (transitional) and limbal — although these clinical divisions are probably not anatomically real (eg Dingledein and Klyce 1989). The central zone usually has an approximately uniform spherical shape but with an irregular periphery and with considerable variability between individuals (Bogan *et al* 1990). The average cornea measures about 11.5mm vertically and 12.1mm horizontally with a typical mean anterior radius of curvature of 7.8mm (Clark 1973b, Ruben 1975, Smith 1977).

Large variations of corneal topography can be expected between healthy individuals (eg Clark 1974, Guillon *et al* 1986, Dingeldein and Klyce 1989, Bogan *et al* 1990, Bores 1991). Topography is affected by endogenous factors such as ethnic race, age, or congenital anomalies (Duke-Elder 1970, Smith 1977, Guillon *et al* 1986, Bores 1991). These may not interfere with vision but are important if any detailed analysis of individual corneas is to be attempted. There are also many external determinants that affect the shape of a cornea and the performance of the eye. These include factors such as the presence of corneal pathology, the effects of surgical intervention, of drug induced changes and of mechanical forces (Duke-Elder 1970, Smith 1977). Probably the most important of these is keratoconus which is a disease that induces severe irregular astigmatism. Surgical procedures to improve defective vision have recently culminated in techniques such as radial keratotomy and photo refractive keratectomy (P.R.K.). The clinical importance of these new refractive surgical procedures is a major impetus to recent research into methods of measuring corneal topography.

Predominantly, because accurate measurement of individual corneas has not been possible, statements about corneal topography have had to be generalisations based on relatively large populations (eg Clark 1974, El Hage 1976a, Kiely *et al* 1982) or models derived from a theoretical consideration of the eye as an optical system (eg Patel *et al* 1993). The relationship between corneal topography and vision is critical, complicated and poorly understood. Quantification of what constitutes normal differences between individuals and normal changes in a single individual is an important step towards understanding corneal abnormalities.

2. KERATOMETRY AND PHOTOKERATOSCOPY

The two most commercially successful approaches to measuring corneal topography rely on measuring the images of mires reflected from the corneal surface. The first of these reflected mire techniques is keratometry, the second is photokeratoscopy (or videokeratoscopy when digital cameras are used). The value of any new method of measuring corneal topography must be established in terms of its benefits over these existing techniques. The principals of keratometry and photokeratoscopy have been addressed in a previous publication (Wise et al 1986) and the problems related to these techniques are only briefly reviewed here. A critical review of these and other methods of measuring corneal curvature is provided by Osborn (1995).

Keratometry, by far the most widely adopted technique, has a number of inadequacies.

- Curvature or net power is presented as a mean value for the central portion of the cornea (typically 3.8 to 4.2mm), but based on measurements at the edge of that area. Over 90% of the corneal surface – the central and peripheral regions – is not measured.
- It is assumed that the curvature of the cornea between two mire reflection points is spherical and thus (for a four point keratometer) that the cornea is ellipsoidal, which may not

be generally the case and is certainly not the case for a keratoconic cornea or irregular astigmatic cornea.

- Keratometer measurements become increasingly difficult when surface irregularities distort the mires.
- The point-by-point nature of the measurement process makes it impractical to compile a complete mapping of corneal topography.

The keratometer fulfils its primary role, that of providing a radius of curvature or, on the basis of an assumed corneal refractive index, a value of the corneal refractive power, for routine clinical assessment of normal human corneas. Its application beyond this role is very limited.

Photokeratoscopes are also, in many respects, inadequate:

- The central and peripheral regions of the cornea cannot normally be measured because of an absence of reflected mires, corneal curvature, and obstruction caused by the nose and brow (Bores 1991, Gormley *et al* 1988, Klyce and Wilson 1989a, Mammone *et al* 1990, Warnicki *et al* 1988). Accurate measurement of the central reflected mire is critical to the mathematical derivation of corneal shape (Missotten 1994).
- The location of reflected mires is highly sensitive to corneal distortion. This is because the position of the reflected mire will be a function of both slope and displacement. On normal corneas, this sensitivity can be advantageous. However when there are very rapid changes in topography the image soon becomes too distorted to measure reliably (Klyce and Wilson 1989b, Friedlander *et al* 1991).
- Visual interpretation of a keratogram can only provide approximate data; clinically significant changes are commonly not detectable (Friedlander *et al* 1991, Wilson and Klyce 1991).
- Several commercial instruments use dithering techniques to fill gaps in the acquired data. The assumptions that are made, typically that the cornea is spherical over any areas that cannot be imaged, are unwarranted (eg Klyce and Wilson 1989a, Bores 1991).
- Photokeratoscopic data are highly sensitive to misalignment with the corneal axis (Heath *et al* 1991) and misjudgment of the focussing position and therefore the distance to the cornea (Saarloos and Constable 1991, Missotten 1994).
- The instruments rely on corneal reflectivity, in turn determined by the condition of the unstable tear film that coats the surface of a healthy cornea and the surface roughness of the corneal epithelium (Duke-Elder 1970). For a healthy cornea, reflectivity is only about 4% at the corneal centre and decreases to near 2% at the periphery (Clark 1973a). Abnormal corneas with low reflectivity cannot be measured (Warnicki *et al* 1988). Photokeratoscopes cannot be used for assessment of corneal topography during surgery because of the inevitable non-reflectivity of the corneal surface.
- Exact topographic data cannot be calculated from a photokeratoscope image. The curve fitting techniques applied are of limited value for several reasons but primarily because of the non-uniqueness of the corneal surface for a given image (eg Wise *et al* 1986, Mammone *et al* 1990), the asphericity of the corneal surface, and because they cannot model abrupt changes that may occur, for example, at the edge of a photorefractive keratectomy (Missotten 1994).

3. THE KERATOCON

The limited quantitative information provided by the keratometer and photokeratoscope together with an increasing requirement in modern ophthalmology for accurate topographic mapping of the entire anterior surface of both normal and abnormal corneas made it apparent that a new instrument was required. It follows from the discussions above that such an instrument should:

i. be capable of measuring the entire cornea,
ii. not rely on corneal reflectance,
iii. not rely on a precorneal tear film,
iv. measure corneal topography with sufficient density and accuracy to provide reliable and clinically interpretable representations of corneal topography and corneal power, and
v. not require that assumptions be made about the geometry of the cornea.

A schematic of the instrument appears below. The cameras used were two 35mm motor-driven Leica R4's fitted with Leica 200mm focal length lenses, positioned at approximately 25° convergence. Because the cameras were non-metric it was necessary to design the prototype so that on-the-job calibrations could be performed. This required photocontrol very close to the cornea — within the limited depth of field of both cameras and sufficiently close to be photogrammetrically reliable. There are two operational considerations. Firstly, the clinician must have access to the eye; secondly, patients cannot be expected to tolerate close proximity to any part of the instrument. To overcome these problems, photocontrol was reflected from a beamsplitter into the optical axis of the system, so that it appeared to surround the cornea in each photograph. A similar technique has been used by Scott (1981, 1987) for his reflex measuring instruments and reported by Mikhail (1974). The control points were a pattern of 30 marks burnt through a thin opaque surface deposited onto the outside of an accurately polished glass sphere. The sphere is flash illuminated from behind at the moment that the cameras are fired. The control points were coordinated to better than ±5μm. The cameras, photocontrol and beamsplitter were mounted onto the platform of a Sun PKS1000 photokeratoscope. This provided a mechanism for controlling the height of the instrument and the position of the patient's head relative to the cameras.

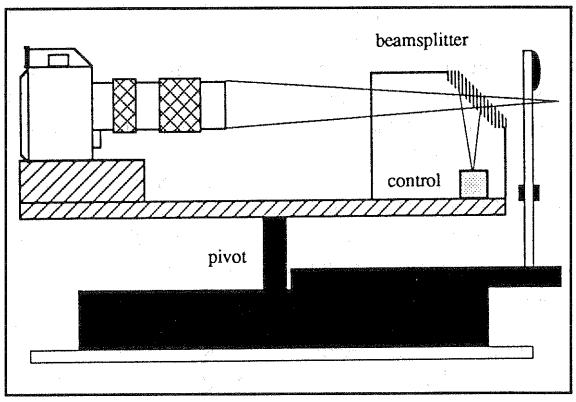

Figure 1. Schematic of the Keratocon.

A variety of methods of marking the cornea were considered including talc, as per Bonnet's work (Bonnet 1959, Le Grand 1961, Bonnet and Cochet 1962) and alternatives to talc, particularly in combination with tear film stabilisers, as well as the self-luminance approaches of El Hage (1972c), Warnicki *et al* (1988) and Banda and Muller (1990). The method adopted uses marked ultra-thin (10μm) hydrogel soft contact lenses. This approach proved satisfactory for the prototype although a teflon material used by Thall (1993, 1993 *per com*) would appear to have greater promise. The indications are that it is simpler to use, it is likely to be simpler to mark, or that a pattern of marks could be projected onto it. The instrument uses two optic fibre bundles to redirect a Mecablitz 45CT flash to the edge of the cornea where much of the light is internally reflected behind the cornea to back illuminate the contact lens. A third optic fibre bundle from the same flash is directed to the back of the control sphere.

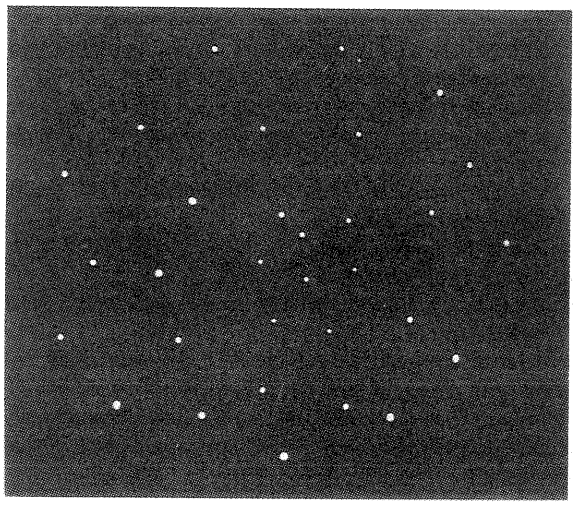

Figure 2. The control points imaged off the beamsplitter in the absence of an illuminated cornea.

The calibration and verification of the system was completed using both the DLT and UNBASC1 software (Abdel-Aziz and Karara 1971, Karara and Abdel-Aziz 1974, Marzan and Karara 1975, El-Hakim *et al* 1979). The DLT was used for routine data reduction. The photocoordinates of the object space control and the corneal reference marks were measured monoscopically using a Zeiss comparator and correlated prior to performing the camera calibration.

It was instructive to make some estimates of the expected accuracy of the system and particularly to examine the likely effects of varying parameters such as convergence angle, base distance and object distance on the expected precision of object space coordinates. The geometry of the system was constrained by minimum object distances, by minimum depth of field considerations, and by the obstructions caused by the patient's brow and nose. The practical working range was found to be an object distance of between 27cm and 39cm, and convergence of between 15° and 30°.

Formulae presented by Abdel-Aziz (1974) and Karara and Abdel-Aziz (1974) were used to investigated the expected precision of the solution for a range of practical camera geometries and the following conclusions drawn:

i. object space precision would be critically affected by the image scale;
ii. optimum precision was likely to be achieved when the camera axes were converging to a point behind the cornea;
iii. for a given image scale, small changes in the camera convergence would have little affect on precision; and
iv. for the best geometry, the object space precision in X, Y and Z would be in the order of 15μm, 15μm and 20μm respectively, leading to an r.m.s. approaching 30μm.

The design of the prototype allowed an experimental evaluation of the accuracy and reliability of the system for a selection of camera geometries. A glass sphere with a radius of approximately 8.5mm was manufactured and its radius measured to an accuracy of better than ±0.5μm using standard optical interference techniques. The (X,Y) coordinates of approximately 20 targets on the spherical surface were measured and corresponding Z-coordinates computed. Their accuracy was estimated to be better than 5μm. Because DLT was the algorithm used in routine data reduction, it was used to photogrammetrically measure object space coordinates for the test points. The measured coordinates of the test points were then transformed to the calibration coordinate system using a least squares three dimensional rigid motion. Accuracy was assessed in terms of the residuals on this transformation.

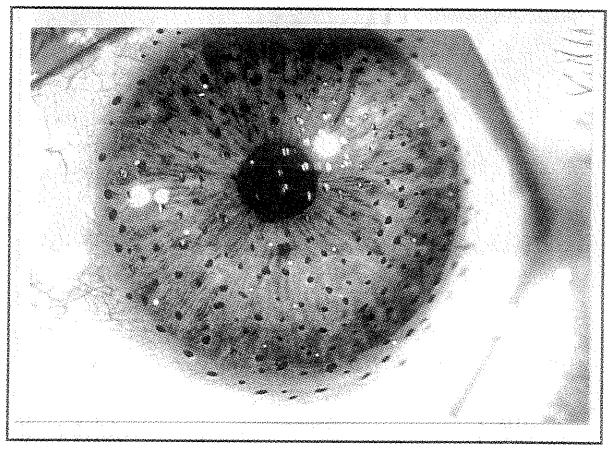

Figure 3. A cornea showing the target points marked onto the surface of a 10μm thick contact lens.

Photogrammetric verification of the Keratocon indicated that the accuracy achieved was consistent with expectations, with the mean magnitude and r.m.s. of object space errors approaching noise levels. Within the working range of the instrument, predicted best precision in X and Y of approximately 15μm (s.d.) compared with measured accuracy of between 3μm and 10μm in X and 4μm and 20μm in Y. Predicted best precision in Z of approximately 20μm (s.d.) compared with measured accuracy of between 6μm and18 μm. The predictions based on the formulae of Abdel-Aziz (1974) were therefore a useful indicator of accuracy.

Clinical verification of the Keratocon proved to be difficult, primarily because there is not a more accurate method of measuring corneal topography and because of the difficulty of testing repeatability. There are two aspects of the instrument's clinical accuracy that still have to be quantified:

i. It is difficult to quantify the accuracy with which the contact lens is representing corneal topography. A

slitlamp was used to visually check the adherence of the lens to the corneal surface for every patient. Comparisons of corneas measured repeatedly and comparisons of Keratocon and photokeratoscope measurements also indicate that the surface of the topography is being accurately represented by the contact lens. The most difficult aspect to measure is the sensitivity of the solution to rapid changes in corneal topography.

ii. The cornea is mapped in a coordinate system defined by the object space control. This means that, although an X-Y-Z translation of the patient's head between successive measurements introduces no errors, it is important that the patient fixates the same point each time the cornea is photographed.

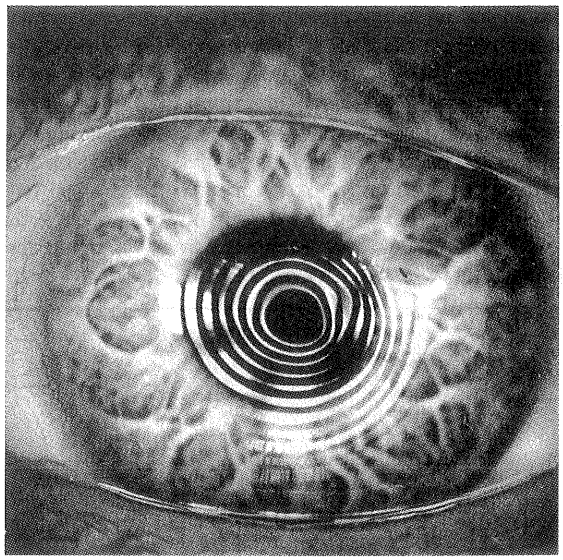

Figure 4. Photokeratoscope image of a patient with conical cornea. Irregular mires can be seen and an impression of downward displacement of the cone. This cornea was too steep to measure with a keratometer.

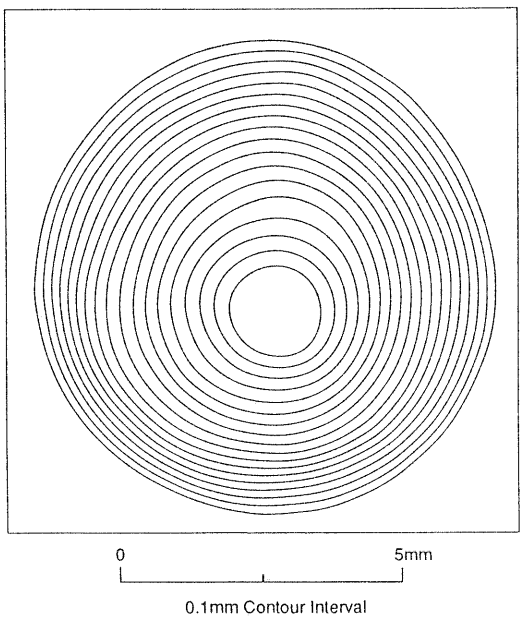

Figure 5. Corneal map showing the displacement of the cone and the flattening of the cornea.

The Keratocon was installed at the Tasmanian Lion's Eye Diagnostic Centre and tested on a selection of patients with corneal abnormalities. Figures 4 and 5 are of a patient with a conical left cornea. The contours show the disturbed topography, known as a "droopy cone". The cornea is flattened above the apex and steeper inferiorly, with temporal displacement of the apex. The photokeratoscope provides very little information about this patient. The cornea was too steep to measure using keratometry. The Keratocon has mapped the cornea to its periphery whereas the photokeratoscope has imaged less than 50% of the corneal surface.

4. A DIGITAL KERATOCON

To meet with clinical acceptance, a biomedical measurement system must satisfy the following conditions:

i. it must not discomfort or intimidate the patient,
ii. it must only require the patient's cooperation for a short period of time,
iii. it must be reliable,
iv. it must output measurement parameters that suit the clinician,
v. it must present those parameters in a readily understandable form,
vi. it must be simple to use,
vii. it must be accurate, and
viii. the time taken to process and present data must suit the application.

The relative importance of each of these conditions will depend upon the application but in broad terms, and certainly in the case of any method of measuring corneal topography, the order in which these have been listed is indicative of their importance.

Accuracy may be the least important of the issues. In the case of corneal measurement this is illustrated by the success of keratoscopic methods. Speed, except in the case of intraoperative procedures, is also not absolutely necessary. The turn around time on clinical procedures such as angiograms and X-rays indicate the delay that clinicians are prepared to tolerate. Ease of use is important, but examination of existing ophthalmic instrumentation such as keratoscopes and slit lamps suggests that clinicians will tolerate quite poor ease of use. The most critical determinants of an instrument's success in a normal clinical environment appear to be the first four on the list, namely the reliable and easily interpreted presentation of appropriate information from an instrument that minimises patient discomfort. Appropriate information includes estimates of how the measured corneal abnormalities will affect vision and the information needed in order that consequent vision defects can be corrected.

A digital prototype of the keratocon should therefore meet the following criteria:

i. sufficient metric reliability to avoid the need for on-the-job calibrations,
ii. real time video imagery for patient alignment,
iii. automatic or semi-automatic target recognition and measurement,
iv. automatic computation of three dimensional data,
v. high data reliability,

vi. high quality presentation of appropriate clinical information,
vii. near real-time presentation of data if intraoperative applications are envisaged.

The investigations reported earlier indicated that image scale is a critical determinant of object space accuracy. For a typical CCD sensor, a 12mm cornea will be imaged at a reduced scale of approximately 1:2, which could be expected to significantly reduce accuracy. This loss may be partly compensated for by the higher object space accuracy that can flow from improved target centroiding, although the relatively poor quality of corneal reference marks is likely to limit such gains, and from the better metric performance of a digital camera, particularly the lack of film deformation. It would be preferable to keep the image scale close to 1:1, which means using a camera with a CCD array that is at least 15mm x 15mm. The choice of image transfer technique is between PLL line synchronised or pixel synchronous framegrabbing from a CCD video camera, or direct capture from a digital transfer camera. The latter two of these techniques provide the best available solution for most photogrammetric applications. An instrument that does not allow the clinician to view the images conveniently in real-time prior to capture would be too cumbersome for routine clinical use. The preferred solution for this application is therefore pixel synchronous framegrabbing of a live video image.

In the case of the Keratocon, and for a 15mm x 15mm array with 1024 x 1024 pixel resolution, an 80μm circular target at scale 1:1 would be imaged onto the array as a region with a diameter of approximately 6 pixels. A target centroiding accuracy in the order of only 0.2 pixel would represent better than ±4μm in the image which is an improvement on the analogue system. Higher resolution arrays would be expected to improve this accuracy.

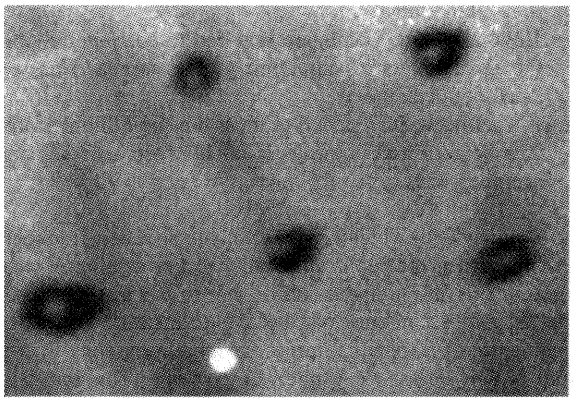

Figure 6. Portion of a 1:1 scale image of a targeted cornea sampled at 3072x2048 pixels.

Target recognition is expected to be the most complicated aspect of a digital implementation. Some experimental work has been conducted in an attempt to ascertain the feasibility of automatically detecting the corneal marks. A portion of a targeted cornea imaged at 1:1 scale on 35mm film and scanned onto a 3072x2048 array is shown in Figure 6 above. The image has been contrast enhanced using a linear histogram remapping but no other image processing. There is reasonable segmentation of the dark marks from the surrounding cornea and within most marks the bright spot representing the target is well enough defined to allow manual measurement. A control point appears as a bright spot and is clearly segmented.

There are a number of factors that complicate automatic recognition. The corneal reference marks are on a contact lens that does not fit in a repeatable manner to the patient's eye and so there is no reliable a priori knowledge of the position of the reference marks in the image. The colour of the iris which forms the background to the corneal images varies greatly between patients. Further, it is difficult to control illumination between images.

If pixel synchronous framegrabbing is used then on-the-job calibrations would not be required. Photocontrol would only be required for initial and regular camera calibrations. The beamsplitter in the analogue instrument could be replaced with a removable mirror that was fitted only for calibrations. The limited depth of field ensures that the cameras would be calibrated for an object space within 1mm to 2mm of the cornea's position. A magnified image of a photocontrol target reflected from the beamsplitter but without a cornea in the field of view is shown in Figure 7. This is again from a 3072x2048 pixel scanned image of the illuminated control sphere. Its diameter is 16 pixels. There has been no image processing used to segment this target from the background. Recognising and centroiding a target such as this would pose no problems. The targets appear in consistent positions in the images and could be automatically identified.

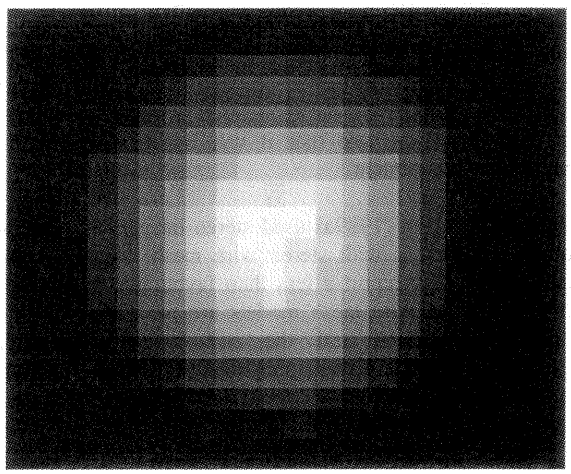

Figure 7. A digital image of a control target imaged off the beamsplitter.

5. DATA PRESENTATION AND ANALYSIS

In medical applications of photogrammetry it is particularly important to address the issues of data presentation and analysis. Unlike most industrial applications, the provision of three dimensional coordinates alone is usually not sufficient. In order for a photogrammetric measurement system to be clinically acceptable, it must provide information that is immediately relevant to the user and in a form that is readily understandable. Different users (the ophthalmologist, corneal surgeon, contact lens fitter, visual scientist) will require different information and different presentation (eg Missotten 1994).

5.1 Visualisation

Clinical diagnosis relies on the human visual capacity to recognise structure and patterns. Effective visualisation is becoming a critical aspect of turning biomedical data into information and is becoming pervasive, particularly for complex decision support applications (eg. Udupa and Herman 1991).

Although commercial instrumentation to measure corneal topography has become increasingly sophisticated, manufacturers have been slow to provide effective visualisation tools. As previously noted the most common visual record, the photokeratoscope image, has limited use. Line drawing representations of the topography such as relief contours (Bonnet and Cochet 1962), wire mesh diagrams (Itoi and Maruyam 1978), departures from sphericity (Clark 1974) and, more recently, colour coded contour maps of corneal surface power (Maguire *et al* 1987, Klyce and Wilson 1989a, Missotten 1994) have improved the clinical value of the information but fall short of ideal because they still do not effectively convey shape information for clinical interpretation (Klyce and Wilson 1989b). More effective presentation schemes are required.

For example, stereographic monitors could be used to visualise the topographic data in three dimensions. The scope of 3D visualisation in medical applications is illustrated by the notion of virtual simulation (eg Rosenman 1991), in which the clinical design is carried out completely on a virtual model. Although technical development and implementation of 3D visualisation in medical instrumentation has tended to lead clinical validation of the real usefulness of the technology (eg Udupa and Herman 1991, Hsu *et al* 1993), its application is rapidly increasing and the indications are that it can significantly improve diagnosis and treatment.

Klyce and Wilson (1989a) used stereo-imagery to illustrate corneal asphericity with stereo wire mesh diagrams. Not surprisingly, they report a less than favourable response by clinicians. The wire mesh models are unsatisfactory three dimensional images and the complexity of mentally interpreting a three dimensional departure from sphericity renders them of little clinical value. Appropriate methods of visualisation need to be developed.

5.2 Surface Matching and Difference Detection

A requirement of the final product is that selected corneal models can be compared in order to:

i. study temporal changes in the corneal topography of an individual patient,
ii. compare preoperative and postoperative corneal topography, and
iii. compare corneal topography of an individual patient with population models or an ideal corneal curvature suited to a particular eye.

The cornea is however an uncooperative surface. The corneal model does not contain any control points and there are no natural targets. Any visible marks such as features on the iris are difficult to identify and geometrically unreliable because they are imaged through a refractive medium. There are no reliable geometric entities such as a corneal apex or corneal edges. Further, the patient's cornea cannot be located in repeatable positions or with accurately repeatable optical fixation. Successive corneal models will not represent exactly the same portion of the corneal surface. Any one may contain only a subset of the others.

Among others, Rosenholm and Torlegård (1988) have investigated DEM matching without control points for absolute orientation of stereomodels in aerial photogrammetry and their methods have been applied in close-range medical photogrammetry by Karras and Petsa (1993). Methods of obtaining a least squares best fit surface match without control have also been investigated by Pilgrim (1988, 1991a, 1991b, 1992) and by Mitchell (1994, 1995). Most methods minimise the difference in separation between surfaces using a least squares solution. Some minimise the angles between surface normals (eg Vezien and Koivunen 1993). In this application, where the clinician may be primarily concerned with corneal curvature and corneal power, it is possible that a matching technique that is sensitive to surface normals will be more appropriate than one which minimises surface separation. Reliable methods of surface matching would significantly improve the utility of any corneal mapping system.

5.3 Derived Quantitative Measures

Corneal curvature is an important derivative of the surface model. Two valuable parameters are a global best fit radius of curvature over the optic cap and a local estimate of corneal curvature at selected points based on data from a limited region. These data are used to calculate corneal power and to characterise different human corneas. The number of points that are required to model the cornea, the accuracy required, and the trade offs between those two parameters have not been determined elsewhere. Questions such as the following must be addressed in order to design a model:

i. How many data points are required in order to compute reliably the radius of curvature of the optic cap along a defined meridian?
ii. How many data points are required to compute reliably the local radius of curvature of the cornea over any small defined region?
iii. In each case, how accurate must the three dimensional data be and what is the trade-off between accuracy and sampling density?

The accuracy with which commercially available instruments measure corneal power is invariably tested using spherical targets. This is unreliable, since any instrument that uses reflected mire techniques will be able to measure the radius of curvature of a spherical target far more accurately than it could a non-symmetric aspherical cornea. The accuracy specifications for a digital implementation of the Keratocon should follow from a proper analysis of the algorithms used to compute parameters such as corneal power and a consideration of the accuracy demanded by current ophthalmic surgery. These issues are now being addressed by some researchers. The indications are that an accuracy in the order of $\pm 5\mu m$ may be necessary for surgical procedures such as photorefractive keratectomy (Missotten 1994).

6. CONCLUSIONS

An accurate knowledge of the topography of the anterior surface of the human cornea is essential in a wide range of ophthalmic applications. None of the commercially available methods of measuring corneal topography provide an accurate topographic map of the entire cornea, particularly when the cornea is diseased or scarred.

A photogrammetric method of measuring corneal topography has been developed. The method does not rely on corneal reflectance and no assumptions need to be made about the approximate or specific geometry of the cornea. Using non-metric cameras, an accuracy of ±20μm has been obtained. The instrument has provided complete and accurate measurements of corneas not well suited to current commercially available corneal mapping systems, particularly grafted and keratoconic corneas.

Full automation of image matching is likely to be the greatest problem to be overcome in a digital implementation of the Keratocon. The task is to obtain sufficient contrast between the targets and the pupil, iris and sclera to allow segmentation of the targets from the background. Additional experimentation with a teflon membrane appears to be the most likely route to a reliable solution.

The ophthalmic market place has demonstrated that many clinicians will be satisfied with an instrument that only maps corneas that have relatively small departures from normality. Given these conditions, and the fact that a photogrammetric instrument is unlikely to be less expensive than a videokeratoscope, it appears unlikely that a digital implementation of the Keratocon is commercially viable if its only advantage is that it provides higher three dimensional accuracy because of its rigorous photogrammetric solution. The key to producing a commercially viable instrument is incorporating into the instrument's software the capacity to apply the three dimensional data to high quality data analysis and visualisation. Existing corneal mapping systems still provide only crude visualisation and analysis tools. A digital Keratocon should provide sophisticated visualisation of corneal topography, be able to derive parameters such as corneal power, and have surface matching capability.

Medical photogrammetry, if it is to be of use to the practitioner, must extend beyond the task of measurement alone. Spatial measurement in isolation is generally of limited value. The task is more likely to be an analysis of temporal change, long term in the case of progressive disease, short term in the case of motion analysis; or a measurement that must be linked to function, such as the functional link between corneal topography and refractive power.

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