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Optical mapping of the thoracoabdominal wall

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ABSTRACT An optical technique has been developed for mapping the size and shape of the thoracoabdominal wall and the change in its shape with breathing. A fixed pattern composed of stripes of light is projected on to both sides of the trunk. These stripes become distorted when viewed from in front and behind, forming contours over the trunk surface. The contours are photographed and then encoded digitally. The digital information can be used to compute automatically the volume of the trunk, the position of any point on its surface, and its cross sectional shape at any level. The technique has been tested on rigid objects (a globe, a cone, and two dummy torsos) that can be measured precisely. With this optical technique linear dimensions can be calculated to within 0·5 mm, cross sectional area to within 5%, and volume to within 1·6–3·7%. These results suggest that this non-invasive technique measures the shape and volume of complex three dimensional surfaces with sufficient accuracy to be tried in clinical practice.

In a previous paper we outlined the principles underlying one optical method of determining the size, shape, and movement of the thoracoabdominal wall.[1] In essence, stripes of light are projected on to both sides of the body and viewed from in front and behind. The illuminated contours of the trunk become obvious at these angles and can be computed automatically. This method is derived from those developed for the face by Lovesey[2] and Cobb[3] and for the trunk by Kovats[4 5] and has some similarities to methods described more recently by Whittle *et al*[6] and Saumarez[7] but, unlike their techniques, it looks at both sides of the trunk simultaneously and lends itself more readily to clinical practice. This paper presents the formal verification of the technique.

Method

APPARATUS AND PROCEDURE

In principle, the patient or the test object is placed in a box of informative light that can be viewed from opposing angles. The patient stands on a platform in a steel reference frame (fig 1) that identifies where the box of light is. The height of the platform is adjusted so that the trunk is illuminated from the larynx to the symphysis pubis. Provided that patients' underclothes are close fitting, they do not interfere with the image, which simplifies the measurement procedure greatly. Patients are positioned in the frame so that the shadows falling between the breasts and between the scapulae are reduced. Usually this requires the patient's coronal plane to be about 45° out of line with the axis of the projectors.

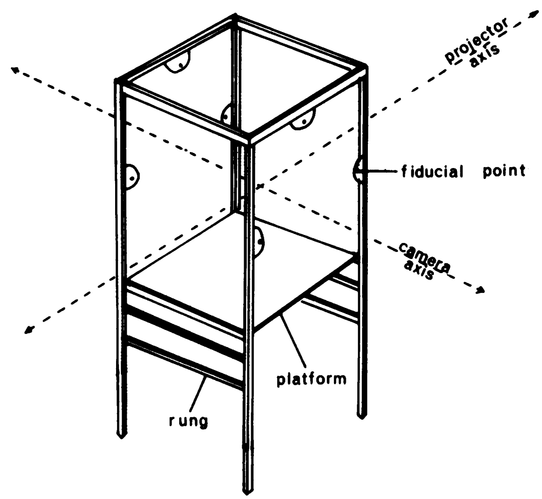

Fig 1 *A diagram of the steel reference frame in which a test object is placed. The patient stands on a platform than can be adjusted for his height. Fiducial points on the frame corresponding to the axes of the cameras and the projectors define the "optical centre" of the system, from which all points are referenced.*

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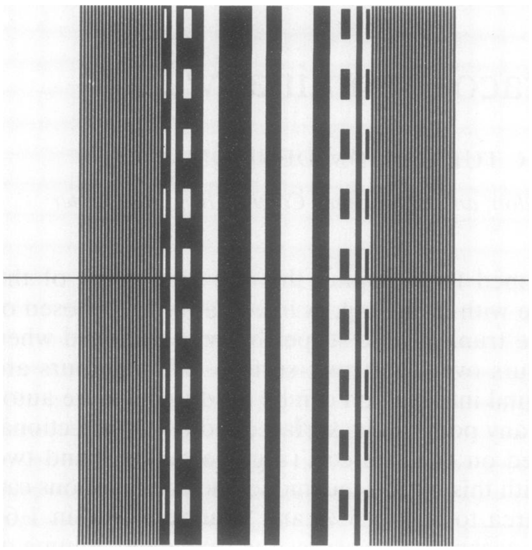


Fig 2 *The projected grid. The vertical black and white lines are of decreasing width and are marked in order to be identifiable on the surface of the trunk.*

The projectors throw a vertical pattern of alternately coloured black and white stripes on to the patient's trunk. The pattern is produced by a 5 × 5 cm perspex slide on to which a grid of lines has been etched by light (fig 2). These lines are so spaced that when they are projected on to an upright cylinder they will divide its circumference into stripes of equal width. Some lines have marks along them to identify them easily. The projectors are arranged opposite each other on either side of the reference frame (fig 3). The distances from the projectors to the frame are not critical, and do not even have to be the same on the two sides, but must be known precisely. The lens and grid combination must then be matched to the distance between projector and frame so that the pattern covers the trunk and creates a sufficient number of contours.

In our case we use Rollei P355 projectors with Heidosmat 85 mm f2·8 lenses placed 1965 mm from the mid-point of the reference frame. The axes of projection are aligned by positioning each grid centre on to the mid-point of the opposite projector lens; we ensure that the stripes are vertical with a plumb bob line. We use grids that produce at least 35 points around the circumference of the trunk at that projection distance. This number of lines is necessary to reconstruct the trunk satisfactorily.

The optical contours are photographed by two cameras mounted opposite each other on a common axis at right angles to but in the same horizontal plane as the common axis of the projectors. The relationship between cameras and projectors is checked by triangulation. The camera to frame distances are not critical but must be known precisely. The camera lenses are chosen so that their field of view just covers the reference frame. Both the cameras and the projectors are focused 100 mm short of the centre of the reference frame, so that anything within remains in the "depths of field" of the optical instruments. This ensures that both the projected patterns and the received images are sharp enough to be acceptable.

We use Olympus OM2 cameras with 50 mm f1·8 lenses situated 1465 mm from the centre of the reference frame. They are positioned with the aid of the split screen focusing device in the camera. The camera is moved until the image of the mid-point of the opposite camera lens and the image of a plumb line hung at the centre of the reference frame are in

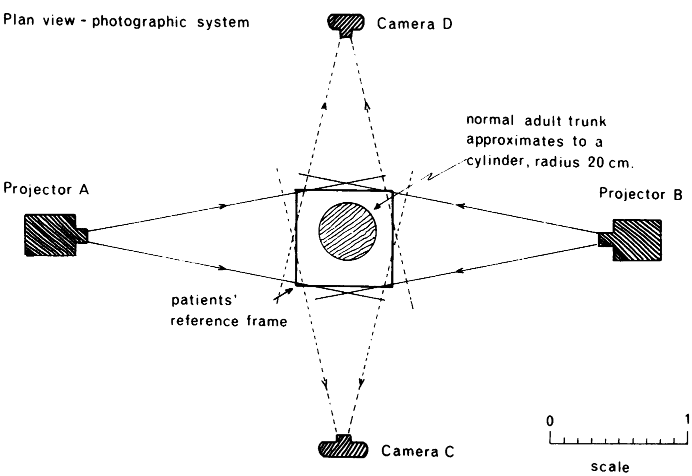


Fig 3 *Plan view of the photographic system. The cameras and projectors are arranged around the reference frame at the same height from the floor.*

line and split lengthwise by the device. At this distance with these lenses the field of view is just large enough to cover anything in the 800 × 1000 mm frame. To capture front and back views of the patient simultaneously the cameras are linked electronically (Control Unit, Olympus M AC). They are wound by motor drives (Olympus motor drive 1) and can be fired at up to 4 frames a second, so dynamic studies can be made during respiratory manoeuvres.

TRANSFORMATION OF THE PRIMARY DATA

To obtain usable information the data on the photographs has to be transformed. One such photograph is shown in figure 4*a*, which in principle is treated as in figure 4*b*—that is, each specific contour line is identified and the true positions in space of many points along it are calculated. For this the position of each point is determined with respect to the intersection of the camera and projector axes, which are indicated by the fiducial points of light on each side of the steel reference frame. The x' and y' dimensions are read off the photograph and the z' dimension is given by the number of the contour line. The grids are calibrated so that the exact positions in space of the vertical planes of light forming the patterns are known. Each contour line formed from a plane of light will therefore have a specific position a known distance from the cameras (the z dimension). This procedure gives the apparent spatial coordinates X', Y', and Z', from which the real coordinates X, Y, and Z are derived by trigonometry. This is necessary since the light beams concerned diverge from the projectors and converge on the cameras. This step produces a table of the X, Y, and Z coordinates of some 500 points on the back and another 500 points on the front of the body, from which the shape of the body surface can be recovered by linear or curvilinear interpolation. Once this surface has been reconstructed, the position of any element, the shape of any cross section, the total surface area, or the volume of all or any part of it can be determined. Details of the calculations are given in Gourlay *et al.*[8]

Clearly, computing the X, Y, and Z coordinates of something like 1000 points has to be automated for the technique to be of practical interest. At present, the still photographs are projected on to a digital plotting table and the X' and Y' values fed directly to a Prime 9100 computer, when the operator runs a cursor down each Z' contour line. We use a Leitz Pradovit projector with a 35 mm lens vertically above a Textronix digitising table. Currently it takes about 30 minutes to digitise each pair of pictures, and a minute or two to transform the data. Soon it will be practical to use videocameras that can digit-

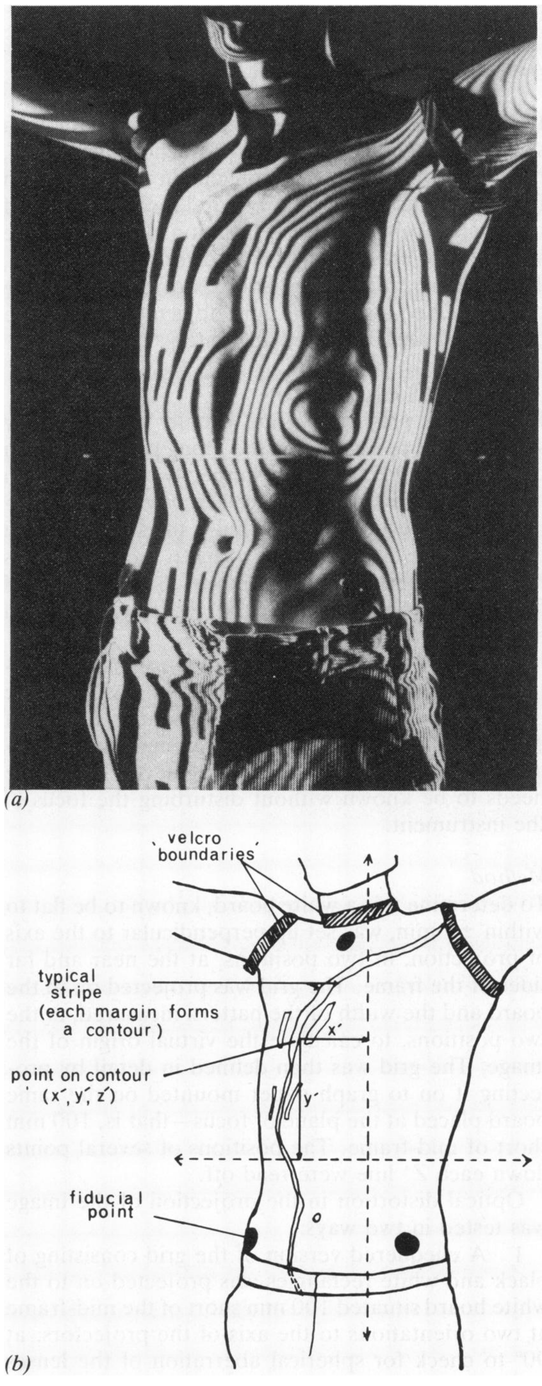


Fig 4 *(a) Front view of a normal subject illuminated from both sides. (b) Diagrammatic representation of (a) showing how the coordinates of one point along a contour are defined. The Velcro tapes and the fiducial points on the trunk define its boundaries.*

ise the images more rapidly.

Assessment of the method

In essence, this method determines body shape by a sequence of four operations: projection of the grid, recording of the pattern, conversion of the pattern from analogue to digital form, and reconstruction of the shape from digital information. Each of those steps contributes to the overall error of the method. To examine the size and sources of error we have photographed a series of rigid objects that could be positioned and measured accurately.

FIDELITY OF PROJECTION

Ideally the projected grids should be absolutely straight, and perfectly focused along planes that can be identified precisely. For simplicity in setting them up they should also be vertical. In practice the etching procedure introduces a finite thickness to each edge, no line is absolutely straight, and the lenses introduce aberrations and limits to the depth of focus. These are amplified by the roughness and angle of the reflecting surface.

First the true planes of divergence of the projected grid must be known. This is not easy because light leaves each projector over the entire surface of its lens and comes into focus only in one perpendicular plane, yet the apparent origin of divergence needs to be known without disturbing the focus of the instrument.

Method

To determine this a white board, known to be flat to within ±1 mm, was set up perpendicular to the axis of projection, in two positions, at the near and far sides of the frame. The grid was projected on to the board and the width of the pattern measured, at the two positions, to calculate the virtual origin of the image. The grid was then defined in detail by projecting it on to graph paper mounted on the same board placed at the plane of focus—that is, 100 mm short of mid-frame. The positions of several points down each Z' line were read off.

Optical distortion in the projection of the image was tested in two ways:

1 A chequered version of the grid consisting of black and white rectangles was projected on to the white board situated 100 mm short of the mid-frame at two orientations to the axis of the projectors: at 90° to check for spherical aberration of the lenses and at 45° to check for angular distortion of the image. The sizes of the rectangles at the edges of the image were measured with vernier calipers to within ±0·1 mm and compared with those expected from the system's geometry.

2 The depth of focus of the lenses was tested by projecting the chequered pattern on to the white board placed perpendicular to the axis of projection at its plane of focus and at 350 mm in front and behind that plane. The image was photographed in each position and the negatives were scanned transversely with an optical microdensitometer. The gradient of light density against distance at each boundary is a measure of the sharpness of focus across the field of view.

Results

Measurements of two sides of each of 40 peripheral rectangles were compared with those expected from the geometry of the projection system and the known grid. When the screen was "face on" to the projector the mean error was ±0·047 mm and the standard error of the mean for these measurements was 0·006 mm. The actual lengths varied from 8 to 44 mm. When the grid image was projected on to a vertical plane at 45° to the projection axis the vertical side lengths of 40 rectangles were on average 0·013 mm shorter than expected, and the standard error of the mean for these measurements was 0·002 mm.

The images created by projecting a chequered pattern on to a white board at the plane of projector focus and at 350 mm in front and behind this plane were photographed and the photographs scanned with a microdensitometer. The average error due to loss of focus across 20 black-white interfaces was 2·2 mm when the board was near to the projector and 2·0 mm when the board was far from the projector.

FIDELITY OF RECORDING

Photographed images of grids projected on to an angled but vertical plane in the reference frame should be absolutely straight, perfectly focused, and "vertical." But the recording lenses introduce further aberrations and limits to the depth of focus. In addition, the quality of the registration of the image depends of the flatness and grain of the film and the efficiency of the developing process.

Method

To test the fidelity of recording, a board 500 × 700 mm (flat to within ±1 mm) was painted with chequered 50 mm squares as precisely as possible. The dimensions of the black and white pattern were measured to ±0·1 mm with vernier calipers. The board was placed vertically in the reference frame at the plane of focus of the cameras—that is, 100 mm short of the middle of the frame—and photographed at 90° and 45° to the axis of the cameras. These photographs were developed and projected on to a perpendicular flat white surface. The accuracy with which the effects of perspective could be corrected was determined by calculating the sizes of the squares at the two sides of the image. These were then compared with the actual dimensions.

To determine the effects of loss of camera focus, the chequered board was photographed perpendicular to the camera axis, at the plane of focus and at 250 mm to either side of this, without adjustment of the camera. The negatives were scanned by the microdensitometer and light density was plotted against distance as before.

Results

When "face on" photographs of a chequerboard were made with the apparatus described for patient mapping, two sides of each of 28 squares were on average 0·15 (SEM 0·097) mm shorter than expected. When the board was oblique to the camera (45°), the vertical sides of the 28 squares were on average 0·16 (0·088) mm shorter than expected. The actual lengths of the sides were 50·03 (0·014) mm.

When the board painted with a chequered pattern was photographed at the plane of focus of a camera and at 250 mm in front and behind this plane and the photographs were scanned with a microdensitometer, the average error due to loss of focus across 10 black-white interfaces was 2·1 mm when the board was near to the camera and 1·95 mm when the board was far from the camera.

FIDELITY OF ANALOGUE TO DIGITAL CONVERSION

An ideal digital plotting system would give the precise location on the image of any point. In fact, errors are introduced in the projection of the image on to the plotting table, in the resolution of the table, and in the ability of the operator to see and follow a contour correctly.

Method

The table and the operator cannot be tested separately so the influence of both factors was tested at the same time, by laying a piece of graph paper over the table and digitising points along a line drawn on the graph paper. Finally, to test the recording and conversion systems together, a photograph of the board with the painted chequered pattern was projected on to the digitising table and the digitised dimensions of several of the squares were compared with those on the original board.

Results

When 28 points on a piece of graph paper, with known X and Y coordinates from +10 to +280 mm, in each sense, were digitised the errors were −0·025 (SEM 0·039) mm for the X value and +0·33 (±0·08) mm for the Y value.

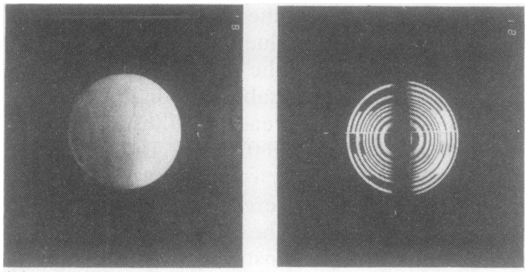
When a photograph of the chequerboard was projected on to the plotting table and digitised, measurements of two sides on each of 100 squares were in error by −0·25 (SEM 0·07) mm. This is a measure of the overall fidelity of the reprojection and digitising process, where no reconstruction algorithms are required. The effect of loss of focus at the limits of the optical instruments was obviously less than the measured 2 mm. This is because the human eye automatically sharpens the images during the digitisation process.

FIDELITY OF RECONSTRUCTION

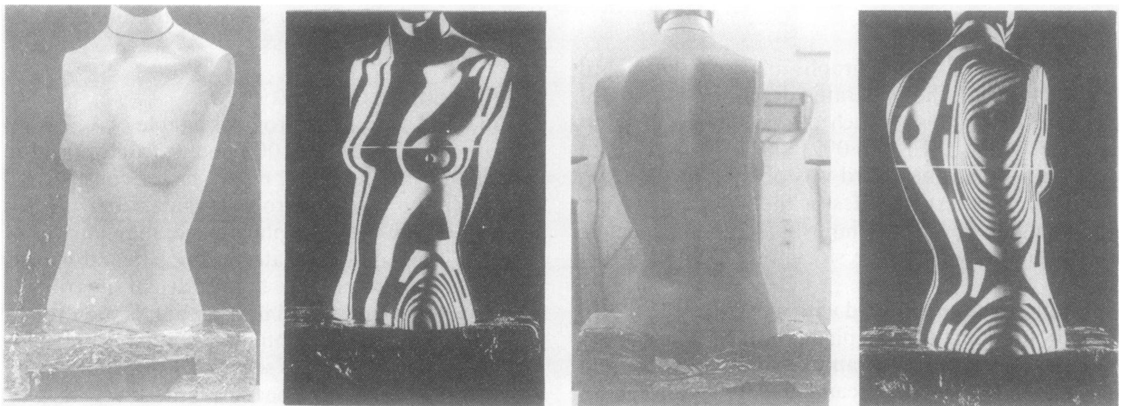
The method eventually provides a table of X, Y, and Z coordinates for about 1000 points, from which the surface of the trunk must be reconstructed. Although the points form a regular array when selected on the photographic image, the matrices of the X, Y, and Z coordinates are positioned irregularly in real space because the surface of the trunk is uneven. Regularly spaced information is required to construct useful descriptions of the trunk—for example, horizontal cross sections. For this the positions of appropriately spaced points must be predicted from the table of fortuitous coordinates, by linear or curvilinear interpolation.

Linear interpolation assumes that the predicted points lie on a flat plane defined by other points. The smallest number of points that will define a plane is three, from which the apparent position of any other point within the triangle can be calculated. The least error in reconstruction of the torso by linear interpolation occurs when the 1000 odd points are linked as triangles to form a multifaceted trunk.

The fidelity of reconstruction also depends on the spatial density of points. For example, to reconstruct a vertical cone by linear interpolation it is necessary to have at least 35 contour lines rising from points around the circumference of its base to calculate its volume to within 0·5%. The volume of the cone can be calculated by considering it in slices; when the thickness of the interval between each slice is known, the volume of each frustum can be calculated from the mean of the radii of the upper and lower faces. Since the human trunk is smoothly curved, curvilinear interpolation is more appropriate and more accurate. We use a spline fitting algorithm which predicts the characteristics of the linking curve by assimilating the information from a short sliding sequence of points in turn. With this form of interpolation the irregularly spaced data are again converted into regularly spaced points that can be used to plot the cross section of the trunk at any level and calculate its volume and surface area.



(a)



(b)

Fig 5 *Views of the globe (a) and of the female mannequin (b) with and without illumination.*

Method

By the use of both linear and curvilinear reconstruction the volumes of two test objects—a globe and a female shop window mannequin (fig 5)—were calculated and compared with the volumes obtained by water displacement of the same objects. With curvilinear reconstruction only, the volumes of two further test objects—a conical waste bin and a male mannequin—were compared with their water displacement volumes. We also compared the areas of optically generated cross sections of the female mannequin with the areas of computed tomography scans at the same levels.

Results

When the globe was reconstructed by linear interpolation from digitised projections of its photograph, 504 measurements of the radius were calculated to average 151·8 (SEM 0·085) mm. The actual radius, calculated from the water displacement volume of the globe, was 151·3 mm. The volume of the globe by water displacement was 14 510 ml. When the volume of the globe was calculated by linear interpolation it was 14 998 ml and by spline fitting it was 14 349 ml. Thus the volume errors are +3·4% and −1·1% respectively.

When the female mannequin was reconstructed by linear interpolation its volume was 18 623 ml, and by spline fitting its mean volume was 16 720 (247) ml. Its water displacement volume was 16 985 ml. Thus the volume errors are +9·6% and −1·6% respectively. Cross sectional reconstructions of the female mannequin produced areas that were 5% (1·9%) smaller at 14 levels by linear interpolation and 5% (0·9%) smaller at 31 levels by curvilinear interpolation than the areas derived from computed tomography. Radiological distortions, however, contribute to these errors to an uncertain extent. The mean volume of the female mannequin by curvilinear reconstruction is the mean of five measurements made in five positions as the mannequin was rotated 10° between each position. To test for reproducibility of this reconstruction, the volumes of the mannequin were calculated from five pairs of photographs taken with the mannequin in one position and also from five digitisations of one pair of photographs with the mannequin in another position. The results were 16 553 (SD 31) ml and 17 179 (16) ml respectively.

When the waste bin, which resembled a truncated cone, was reconstructed by curvilinear interpolation its volume was 19 458 ml. Its water displacement volume was 18 994 ml. Thus the volume error was +2·4%. When the bin was photographed in each of four positions within the reference frame its mean volume was 19 696 (256) ml. The mean volume error was therefore +3·7% (3·70%).

When the male mannequin was reconstructed by curvilinear interpolation its volume was 23 030 ml. Its water displacement volume was 22 575 ml. Thus the volume error was +2%.

Discussion

The results indicate that the overall fidelity of reconstruction of complex curved shapes is within about 0·5 mm for linear dimensions and within ±3·7% for curvilinear reconstruction of volume, with good reproducibility. This in turn suggests that the method of optical contour mapping described here is sufficiently accurate for many clinical purposes. It is therefore reasonable to ask how it compares with other methods of studying the dimensions of the trunk, and what functional information it is likely to provide.

The shape and movement of the thoracoabdominal wall can be determined by various methods—for example, those using magnetometers or inductance plethysmography—that provide information in two dimensions only. For three dimensional measurement, procedures such as vector stereography[9] have been used but, like magnetometry and inductance plethysmography, they require direct contact with the trunk surface, which may disturb the shape or motion of the wall.

Optical techniques do not require such contact and can be very accurate. They include light sectioning,[10] holography,[11-13] stereo photogrammetry,[14-17] moire interferometry,[18-23] and light projection.[1-6 24] The first of these, the projection of a single plane of light on to the trunk, is cheap and simple but it allows observation of only one section of the trunk at a time. Holography does not have this limitation, but it is expensive to set up. Stereo photogrammetry, the comparison of two photographs of the trunk taken from different angles, is cheap and accurate but must be done by hand and eyes, which precludes the development of an automated system. Moire interferometry, which is the generation of interference fringes by projecting light through a grating and observing the trunk with a camera offset from the projector, has been used successfully by many workers concerned with shape measurement; but the contours produced tend to have ill defined edges, making automatic contour detection difficult. Some engineering and computing aspects of the procedure have been reviewed recently by Duncan and Mair.[25]

We chose light projection because the technique is simple and safe and uses domestic equipment that is cheap and widely available, and becaue the patterns produced cover the whole surface of the thoracoabdominal wall. These patterns can be read and measured easily, which means that a fully or almost fully automated system for studying the thoracoabdominal wall both at rest and in motion is a reasonable objective.

The results obtained when the mannequin was rotated or the waste bin was translocated indicate that the error in computation of trunk volume varies with the angle of illumination. It is likely that the major loss of accuracy occurs in the reconstruction. This may be because, in some positions, too much of the trunk was in shadow (areas of uncertainty) or an insufficient number of sampling points were taken from the illuminated areas. Shadow can be reduced either by using more projectors[7] or, as we do, by adjusting the position of the object so that valleys are illuminated. The number of contour lines falling on the object depends on the lens and grid combination used. Either the lenses or the grids can be changed. In many measurements of respiratory function the change in trunk volume is more important than its actual size, so that systematic error introduced by the reconstruction algorithm is often cancelled out.

What can be gained from precise reconstruction of trunk shape and volume and the change in these variables with respiratory manoeuvres? If the volume of the trunk could be calculated at the beginning and end of a breath, before and after blowing into a manometer at a predetermined pressure, and at the beginning and end of a timed breath hold after inhalation of a soluble gas such as nitrous oxide, it would be possible to measure respired volume, compressible gas volume, and effective pulmonary blood flow respectively. In addition, the spatial distribution of each breath could be determined to assess the relative contributions of rib cage and abdomen to breathing. Measurement of the shape and change in shape of the thoracoabdominal wall can be used to monitor the progress and define the effects of treatment in disorders which affect the chest wall, such as scoliosis, poliomyelitis, and spinal injury. Other disorders such as asthma, emphysema, and fibrosing alveolitis also affect chest wall mechanics and it may now be possible to study this aspect of these diseases more easily. Recently we have modified the present technique for use in bed bound patients who cannot be seen "in the round." The modified technique appears to be as accurate as the procedure described here.

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