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Influence of the distribution of stiffness in the human left ventricular myocardium on shape change in diastole

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Abstract—*The finite-element method of stress analysis was applied, in its 3-dimensional form, to models of the left ventricular wall of two patients. The basic geometry and pressure changes were available from biplane angiographic and catheter data. A parametric study of the influence of the distribution of assumed wall stiffness was carried out on the computer models. By comparison with the experimental data the significance of the various changes could be assessed. For example, stiffness of the annulus fibrosus appears to exert a major influence on the deformation and volume change of the ventricle whereas that of the apex does not. The presence of infarcts in the region of the base are more significant than those near the apex. The spiral structure of the myocardium, and the ratio of longitudinal/circumferential stiffness are major determinants of diastolic wall position. These results appear to have appreciable clinical significance.*

Keywords—*Diastolic shape change, Left ventricular myocardium, Stiffness*

1 Introduction

THERE IS A great deal of evidence to suggest that abnormalities of ventricular filling can interfere with overall cardiac function. In the presence of left ventricular hypertrophy, as assessed from e.c.g. or increased left ventricular muscle mass, left ventricular pressure-volume or stress-strain relationships are abnormal, due to an altered modulus of elasticity (GIBSON and BROWN, 1974; MIRSKY, 1976).

Viewed from an engineering standpoint, the myocardium of the left ventricle can be regarded as a thick-walled pressure vessel of irregular shape. During diastole, the major factors affecting its behaviour are the intraventricular pressure, the geometry of the wall, the distribution of thickness and the properties of the myocardium itself. To carry out a mechanical stress analysis with reasonable accuracy therefore requires a knowledge of these basic variables. As the ventricle is a 3-dimensional object without any special planes of symmetry, profiles in at least two planes are needed at each instant. This is most satisfactorily provided by biplane angiography (with computer processing of the resulting images). Pressure is reliably measured with a catheter-tipped manometer, and wall thickness either by angiography or M-mode echocardiography, the latter method being the more satisfactory but applicable only to a single region. The material of the myocardium is both nonhomogeneous and anisotropic. Further, it is generally considered that the behaviour of cardiac muscle is not simply Hookean, but also that it exhibits viscoelastic properties. Values for these properties, either from tests on cadaveric material (LUNDIN, 1944) or from direct measurement *in vivo*, have been proposed.

Attempts by engineers and cardiologists to perform stress analysis on the left ventricle were initially based on the assumption of a linear elasticity and began with rather unrealistic geometrical models, such as spheres or ellipsoids (SANDLER and DODGE, 1963). In developing more satisfactory methods, an important advance has been the application of the finite-element technique (ZIENKIEWICZ, 1973), which possesses distinct advantages over the other methods of stress analysis. Compared with classical analytical techniques, it is able to model much more closely structures of irregular geometry, and it is able to take into account nonhomogeneous and anisotropic material properties. It is therefore particularly suitable for myocardial stress analysis. Its use makes it possible to describe and therefore to investigate alternative models of the left ventricular wall which accord more closely with anatomical observations.

2 Methods and typical application

The technique described above has been used to model the passive mechanical/structural behaviour

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of the ventricles of two patients, at various stages in the cardiac cycle, simultaneous intra-ventricular pressure values being available from a pressure transducer at the tip of the catheter. Its application will be illustrated by considering one increment in time (20 ms) during the diastolic recording for one patient, denoted here as N. Digitised anterior-posterior and lateral outlines of the endocardium were available and are those shown in Fig. 1 for a time increment between frames 1 and 2 of 0·02 s at the point of minimum pressure. The increment of intraventricular pressure over this period was 0·133 kN/m^2 (1 mmHg). Fig. 1 also shows the small sectors of epicardial profile recorded on the radiographs. Using the geometric modelling procedure described by YETTRAM and VINSON (1979*b*), the three dimensional shape was developed. The wall thickness is clearly based on scant data, but from that available the epicardial profile was fitted to give an apical vortex diameter of 10 mm and thickness of 2 mm as suggested by KEITH (1907). The finite-element model consisted of 36 3-dimensional 20-node, isoparametric elements arranged in six layers of six elements per layer, as shown schematically in Fig. 2.

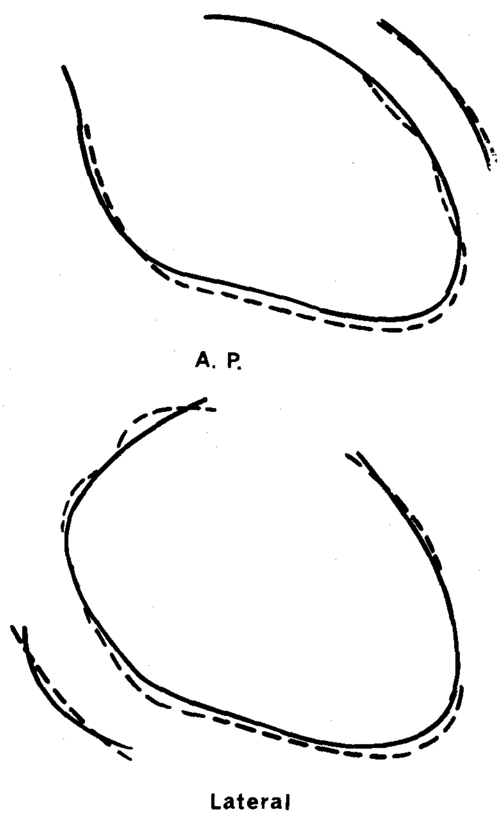

Fig. 1 Digitised biplane outlines from cine-angiocardiographs
——— First position
– – – – Second position

The first layer of elements was only 1 mm deep to simulate the annulus fibrosus, ZIMMERMAN (1966).

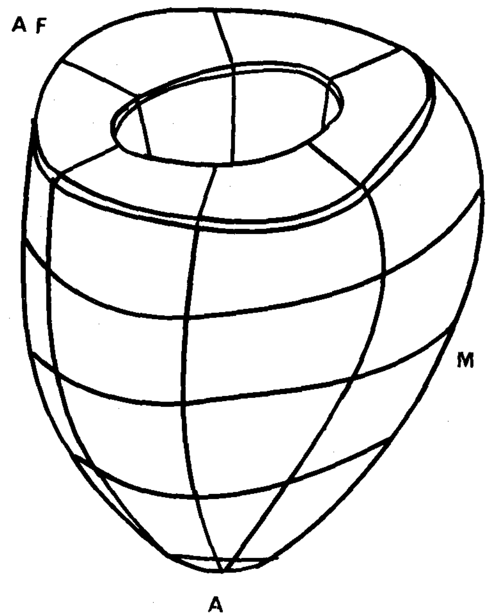

Fig. 2 Typical 3-dimensional finite-element mesh for left ventricle
AF—annulus fibrosus layer; M—body of myocardium; A—apex

There is no definitive information on ventricular constraint in the literature. The kinematic restraint used here is based on clinical observation. The model was constrained at all the nodes along the inner periphery of the base in the vertical direction and completely restrained at three points adjacent to the septum. The increment of diastolic pressure was applied by means of consistent, uniform normal pressures at the endocardial nodes.

As yet there is no definitive method available for obtaining the specific comprehensive set of 3-dimensional mechanical properties of a particular ventricle.

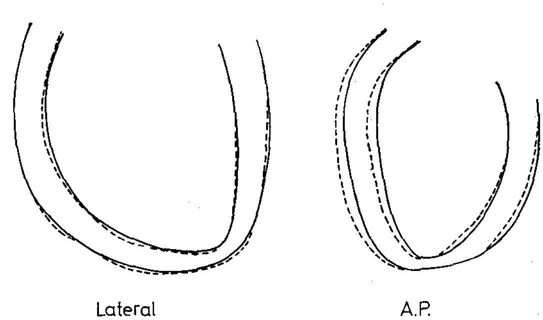

Fig. 3 Deformation of finite element model of left ventricle.
——— Initial position
– – – – Deformed position
Elastic properties used as in Table 1

However, here the elements of the assemblage were given various combinations of elastic properties and the resulting deflection patterns examined and compared with the observed profiles from the cineangiographs. A review of suggested values for these properties, collated from the work of various workers, has been presented by YETTRAM and VINSON (1979*a*). A specimen set of properties is given in Table 1. The annulus fibrosus is considered as isotropic, whereas the major part of the myocardium is orthotropic with the longitudinal stiffness one-half of the stiffnesses in the meridional and crosswall directions. KEITH (1907) has suggested that the apex consists of stiff fibrous material, and therefore an extremely high value of modulus of elasticity was chosen for this region.

The resulting deflection profiles from this analysis are shown in Fig. 3. These are based on the displacements of the nodal points on midplane sections taken to represent the anterior-posterior and lateral views. It should be noted that changes in the outlines of these two profiles cannot completely represent the movement of the left ventricle as a whole. For

Table 1. Elastic moduli used for finite-element analysis

E_x	8	G_{xy}	5·41	μ_{xy}	0·24	μ_{yx}	0·48
E_y	16	G_{yz}	5·41	μ_{yz}	0·48	μ_{zy}	0·48
E_z	16	G_{zx}	5·41	μ_{zx}	0·48	μ_{xz}	0·24

E = modulus of elasticity, μ = Poisson's ratio, G = modulus of rigidity, x = longitudinal direction, y = meridional direction, z = crosswall direction. Units for E and G are kN/m²

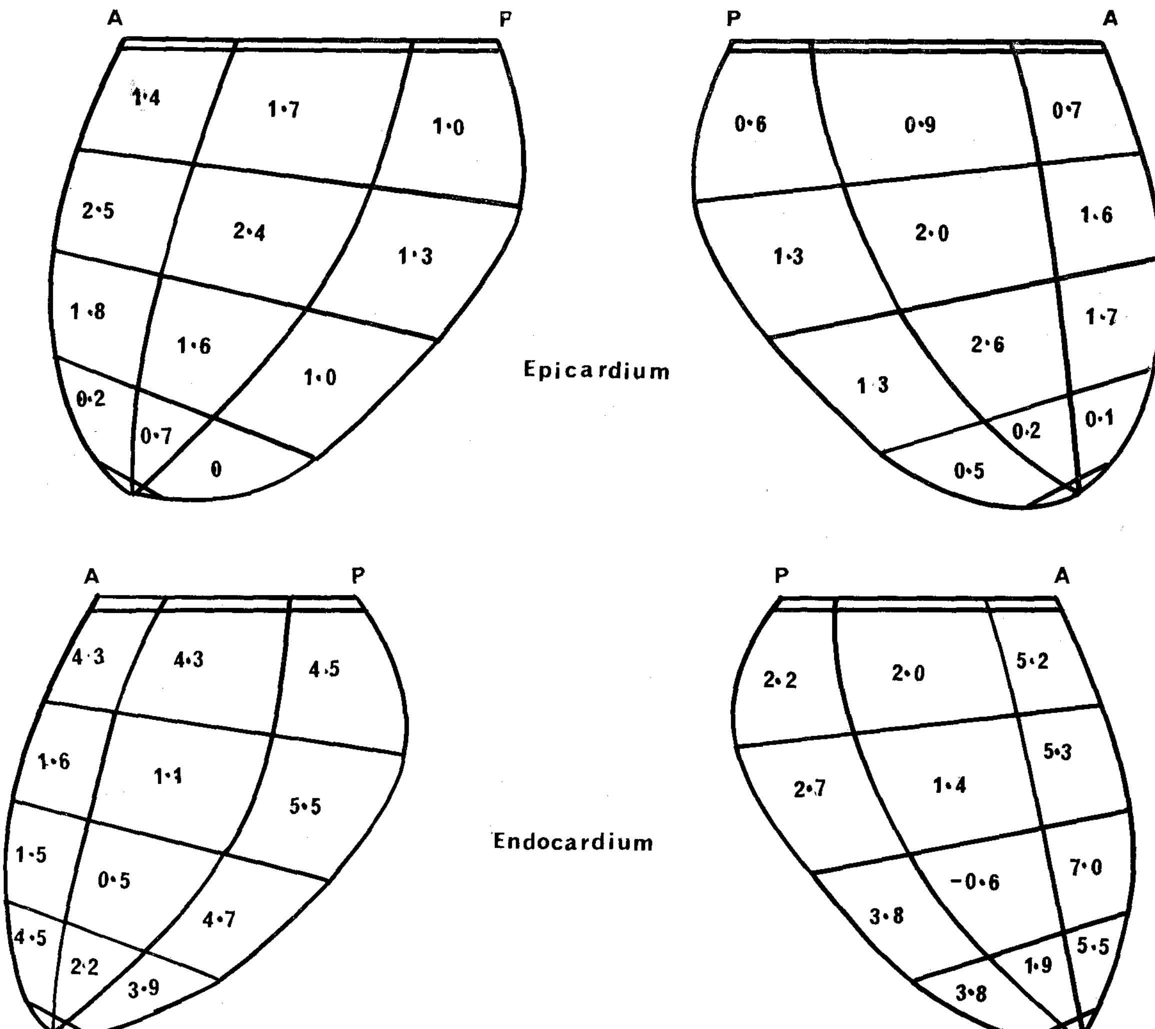

Fig. 4 Patterns of the maximum absolute values of principal direct stresses per unit internal pressure over epicardial and endocardial surfaces

example, it is possible for regions away from the plotted sections to deflect so as to form a larger periphery. Therefore the displacements shown in Fig. 3 are nodal point displacements and are not necessarily equivalent to those given, essentially from X-ray silhouettes, in Fig. 1, and caution must be exercised in comparing the two figures.

Fig. 4 shows the distributions of the maximum principal stress, per unit internal pressure, on the epicardial and endocardial surfaces, for the particular case being examined. The highest values appear to be over the right anterior (septal) and left posterior regions of the endocardium. The values obtained compare generally in order of magnitude with those of HAMID and GHISTA (1974), although their highest values appear in the apical region.

It is not possible to obtain exact, comprehensive deflection data which describe the complete deformation of the left ventricle. However, the purpose of this work is not specifically to compare analytical and observed profiles but to detect trends in behaviour for variations in stiffness parameters.

3 Parametric studies of left ventricular behaviour

Two patients were studied, denoted by M and N. Patient M had been investigated for chest pain but the left ventricular angiogram and coronary arteriograms were considered normal. Patient N also showed a normal coronary arteriogram but had a moderate left ventricular dilation of unknown cause. Although the wall thickness in N was greater than in M, both are within the normal range. Haemodynamic results are given in Table 2.

Various changes were made to the distribution, form and values of the various elastic moduli assigned to the elements of the mathematical model. From an examination and comparison of the resulting displacements and stress distributions certain broad biomechanical conclusions can be drawn.

3.1 *The annulus fibrosus*

As previously suggested (ZIMMERMAN, 1966; JANZ and GRIMM, 1972), the annulus fibrosus was initially modelled as a ring of collageneous tissue at the base of the ventricle. Values for the modulus of elasticity for collagen have been suggested by many investigators and range from 2×10^5 to $1 \cdot 15 \times 10^8$ kN/m^2. This range indicates a material very much stiffer than that of the remainder of the myocardium for which values of modulus of elasticity have been proposed lying between $0 \cdot 5$ and 1×10^3 kN/m^2.

Using a typical value for the modulus of elasticity for the annulus fibrosus [$3 \cdot 9 \times 10^5$ kN/m^2 (GRATZ and BLACKBERG, 1935)] and that for the remainder of the myocardium as 16 kN/m^2 for patient N and as 64 kN/m^2 for M, analyses were carried out for both patients. Deflection profiles were plotted and compared with the movements recorded by cine-angiocardiography. In the anterior-posterior view the model predicts an overall 'radial' expansion rather than a sideways displacement plus expansion of the left ventricle shown by the angio recording. The lateral view showed no significant displacements.

When the annulus fibrosus was then remodelled for both patients, as having the same material properties as the myocardium, the sideways movement, towards the septum, observed in the anterior-posterior view was correctly predicted in both. These results then would not substantiate the modelling of the annulus fibrosus as a ring of collagen.

It would appear therefore that a stiff annulus fibrosus stops the outer edge of the base of the ventricle from moving upwards, which in turn stiffens the whole ventricle, and greatly reduces the displacement toward the septum observed in the anterior-posterior view. Further, the overall increase in volume is less than that observed or that predicted with a flexible annulus fibrosus. The overall movement towards the septum would then appear to be linked with an advantageous volume increase. The fact that the annulus fibrosus is flexible and does deform has been observed by LIAN (1909).

Table 2. Haemodynamic values

	Case 1	Case 2
Right Atrial Pressure, mmHg	$\overline{3}$	$\overline{19}$
Right Ventricular Pressure, mmHg	25/0, 5	57/10, 19
Pulmonary Artery Pressure, mmHg	25/10($\overline{13}$)	55/23($\overline{30}$)
Left Ventricular Pressure, mmHg	120/5, 10	130/12, 25
Cardiac Index, $l min^{-1} m^{-2}$	3·6	3·0
Ejection fraction, %	68	45

3.2 *The apex*

KEITH (1907) has suggested that the apex is a small thin area of stiff fibrous material formed as a vortex of the ends of the spirally arranged fibres. It seems reasonable to represent this area as a small 'button', about 1 cm in diameter, of collagen.

Comparison was also made with the apex having elastic properties equal to, and a factor of 10 and 100 times greater than, the myocardial value suggested for diastole. There was no significant difference in the predicted deflections for either M or N for these tests. The apex stiffness appears to have no influence on the deflection of the ventricle.

3.3 *The myocardium—isotropic and homogeneous*

Initially the values of modulus of elasticity (E) for the models were chosen from earlier ellipsoidal analyses (YETTRAM and VINSON, 1979*a*). Model M was given a value of E of 64 kN/m^2 and model N of 16 kN/m^2, Poisson's ratio was taken as 0·48. The shear modulus (G) was found from the isotropic relationship as 21·6 kN/m^2 and 5·41 kN/m^2, respectively. In each case the mitral annulus was given the same properties as the myocardium.

The prediction for N gives displacements of the correct order of magnitude compared with the biplane. M is much more difficult to interpret.

The biplane for M shows that towards the apex an inward movement is observed on both sides of the anterior-posterior view, while towards the base of that view no movement at all occurred. This seems incompatible with any passive displacement pattern. In M the pressure after the minimum pressure point was changing rapidly: eight times faster than N. It may be that, although the angiographic outline at minimum pressure was taken, the inward movement is the early onset of active tension development and it is the start of the upward moving contractile wave which is being observed. The displacements observed for M do indicate that the value of E must be at least the value used of 64 kN/m^2. This means that the myocardium of M was significantly stiffer than that of N.

3.4 *The myocardium—isotropic and non-homogeneous*

Following the suggestion by NIKRAVESH (1976) the ventricle of M was modelled as being stiffer towards the base in approximately the same manner as he had used. The elastic modulus for each of the four layers of myocardium were represented as 128 kN/m^2, 64 kN/m^2, 32 kN/m^2 and 16 kN/m^2, respectively from base to apex. The value of G was calculated from these using the isotropic relationship. The resulting deflections did not improve the modelling. In general terms this representation allows too much expansion to occur in the flexible apical end of the ventricle.

As the region near the apex of the model M actually appears to move inwards, the above sequence of material properties was reversed so that the ventricle was stiffer at the apex than it was at the base. The resulting deflections are far too large. They allow too great an expansion of the ventricle as a whole.

The values of E used in these analyses are three to 11 times larger than values, of the order of 5 kN/m^2, normally suggested for this period in the cardiac cycle. However, the earlier work on ellipsoids, and the similar work by GHISTA *et al.* (1975), do show, as was mentioned earlier, a wide variation in the possible values of E.

3.5 *The myocardium—crosswall stiffness*

JANZ and GRIMM (1972), from a study of rat ventricular muscle, have suggested that the transverse or crosswall value of E in the outer two-thirds of the ventricle is one-half of their suggested tangential value. Their value for shear modulus was found to be numerically one-quarter of the tangential value of E. To enable the effect of the transverse modulus to be assessed the myocardium of model N was given the following properties: tangential E of 16 kN/m^2, transverse E of 8 kN/m^2, G of 5·41 kN/m^2. Hence the transverse modulus is halved but the shear modulus is kept at the previous value for isotropy. The angles of the local property directions to the global directions were estimated from the model. The maximum change in displacements at any one point are found to be only slightly larger than those predicted by isotropy. It should also be noted that wall thinning was not increased by the change in modulus. Thus it would appear that the ventricle is not particularly sensitive to the value of the transverse modulus, remembering that the values being used here are higher than those normally suggested. However, it does seem logical that the transverse direction, across the fibres, should have different material characteristics to the tangential direction in the plane of the fibres.

3.6 *The myocardium—variation of shear modulus*

Several previous investigators have de-emphasised the effect of changing the shear modulus (NECKYFAROW and PERLMAN, 1976). However, changing the shear modulus from 5·41 kN/m^2 (from isotropy) to 4 kN/m^2 causes a pronounced increase in deflections, and with the value for the shear modulus of 1·52 kN/m^2 suggested by JANZ and GRIMM (1972), the overall deflections are greatly increased in the anterior-posterior view and to a lesser extent in the lateral view. It would appear from this that the value of the shear modulus can in fact greatly influence the displacement characteristics of the left ventricle.

3.7 *The myocardium—longitudinal/circumferential stiffness*

Earlier work on the ellipsoidal models has suggested that the ratio of the longitudinal to the circumferential elastic modulus was 0·91 (YETTRAM and VINSON, 1979*a*). Implementation of this produced no significant change from the displacements as predicted by an isotropic relationship. It would appear from the more realistic 3-dimensional approach which we have used that excessive sensitivity to this ratio is merely a function of an ellipsoidal model.

A further reduction in the ratio to 0·5, however, would allow slightly too great an elongation and a general expansion of the ventricle (Fig. 3). This indicates agreement with the results of NECKYFAROW and PERLMAN (1976), who also suggested from a thick-walled-shell finite-element model, but based on single-plane angiocardiography and assumed circular cross-sections, that the ratio for the best fit for displacements was 0·8.

3.8 *The myocardium—effect of fibre angle*

LOWER (1669) has observed a fibre angle change through the myocardium of $+60°$ to $-60°$. If this is taken as approximate and the inner third is considered to be more flexible (JANZ and GRIMM, 1972), then there may well be an overall effective spiral angle for the fibres away from the longitudinal and circumferential directions.

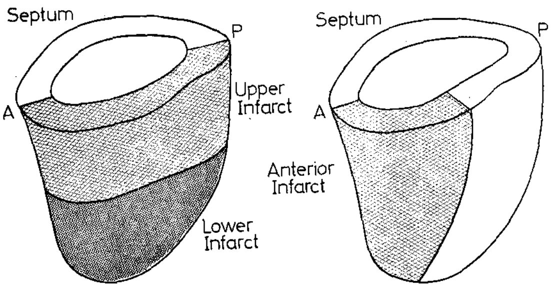

Fig. 5 Extents of three separate infarct cases studied

Two offsets of the longitudinal fibres for models M and N were examined. A spiral arrangement of these fibres of 30° and 10°, clockwise from base to apex was chosen. The spiral effectively makes the wall less stiff, thus giving increased longitudinal displacements. In the lateral view significant displacements now occur. However the overall displacement towards the septal side in the anterior-posterior view is reduced.

It is possible to infer from these results that an overall offset of the fibre angle would allow easier passive expansion in diastole. However, although correct representation of the lateral view is aided, this is not so for the anterior-posterior view. The overall effect that the fibre direction may have on left-ventricle function appears not to have been

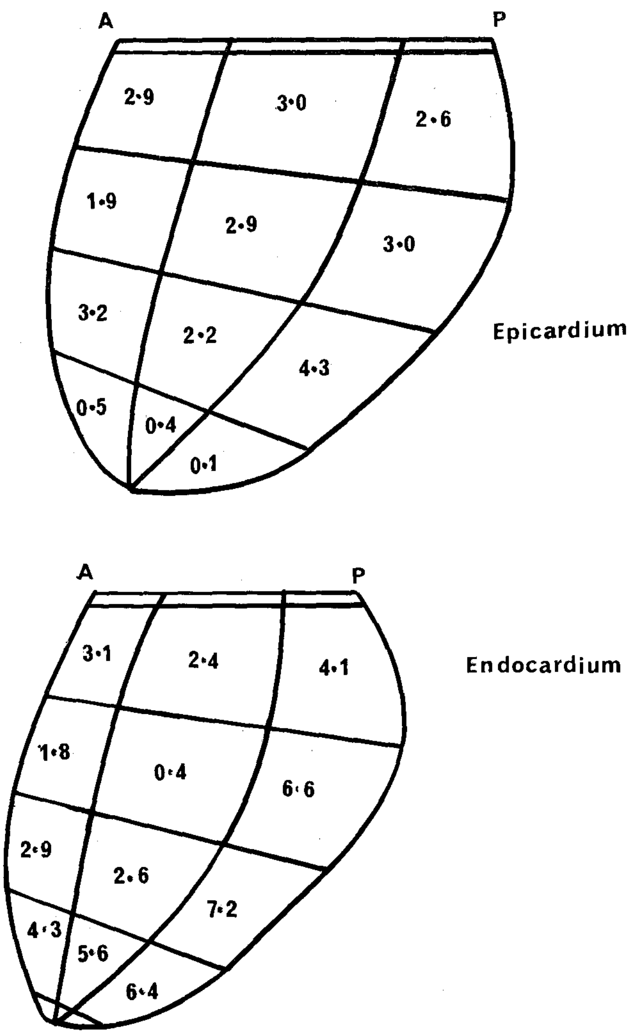

Fig. 6 Patterns of the maximum absolute values of principal direct stress—Patient M

considered in the recent literature, yet the result of STREETER (1969) and many other workers show the complex variation of fibre angle throughout the wall.

3.9 *The myocardium—the effects of infarcts*

Infarcts initially are soft areas of necrotic tissue within the myocardium. After two to four weeks the tissue hardens and becomes fibrotic. The models were therefore tested with both stiff and flexible areas to simulate infarcts.

Three infarcted regions were chosen, each covering 25% to 30% of the myocardium of model M which might be expected seriously to interfere with cardiac function (PARMLEY *et al*, 1973). The first and second infarcts are the upper and lower halves of the outer wall, respectively, and the third is an anteriorly positioned vertical strip (Fig. 5). The thinning to about 80%, generally observed at an infarct, was not simulated.

The predicted displacements for the flexible isotropic infarcts, with the modulus of elasticity equal to one-quarter of the myocardial value, show mainly distension in the region of infarction. However, the infarct at the base of the wall has a greater effect on wall movement towards the apex than in the opposite case. The stiff infarcts were given a value of elastic modulus equivalent to that of collagen, quoted earlier. The ventricle is thus modelled as effectively carrying a passive region of stiff tissue. Of the stiff infarcts the two at the base of the ventricle have the most effect. These appear to stiffen not only the infarcted area but also the whole ventricle. The resultant disturbance of movement is not generalised or localised to the infarcted area but abnormal movements appear both at the infarction itself and in the opposite wall. Thus restriction of movement need not necessarily indicate an infarcted

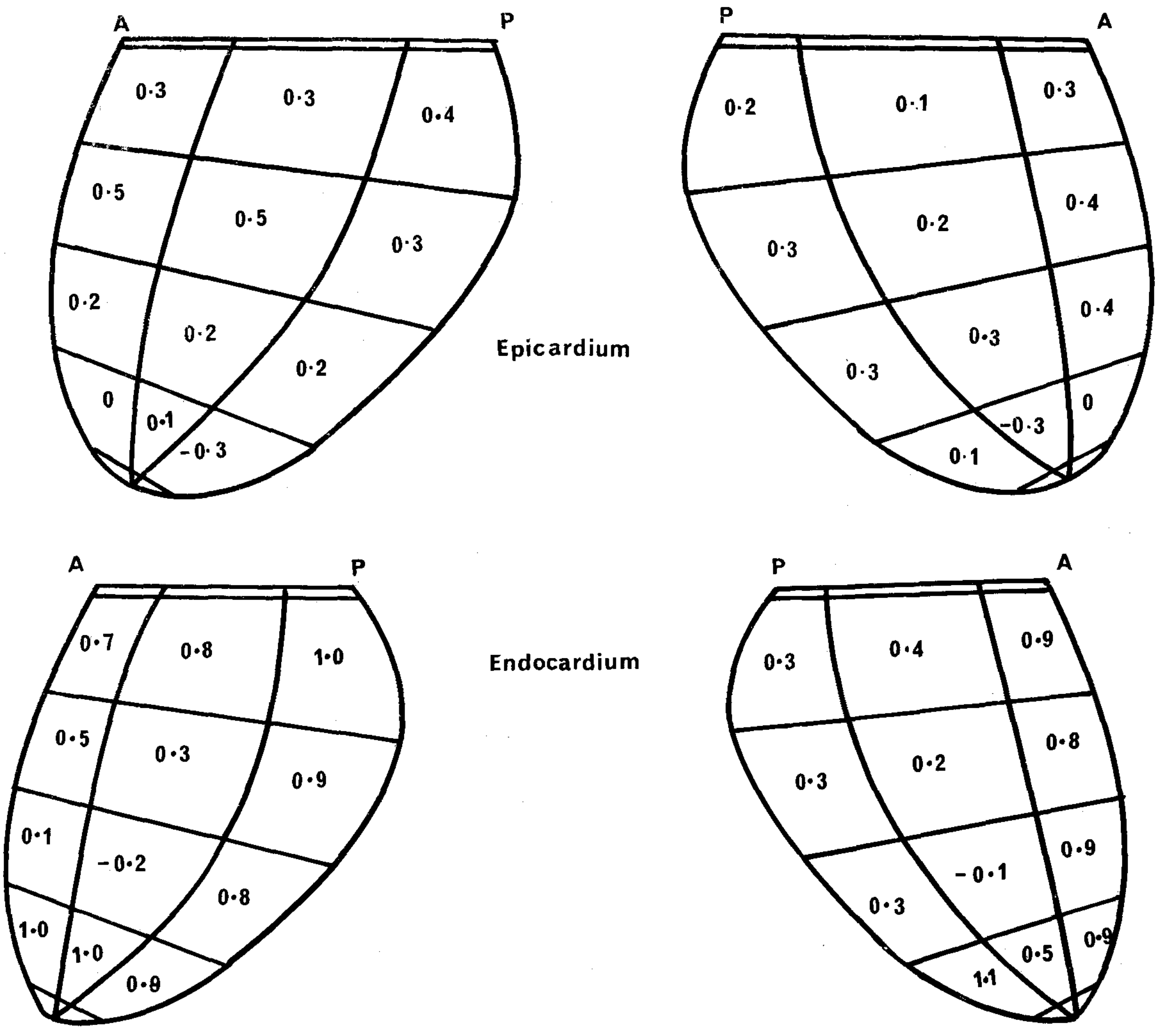

Fig. 7 Patterns of the maximum absolute values of principal direct stress—Patient N

region but may be the reflection of such a region elsewhere. By contrast, the apical infarct showed only a local effect and did not appear to interfere to the same extent with the movement of the ventricle as a whole.

3.10 *The myocardium—effect of right ventricular pressure*

Accurate modelling of the left ventricle would seem to require that consideration be given to right ventricular pressure. In diastole the normal right ventricular pressure is less than that in the left ventricle. However, elevation of right ventricular diastolic pressure may be associated with deviation of the septum and distortion of the left ventricular cavity (SANTAMORE, 1976). It is assumed in this study that as a total system of forces the right ventricle does not influence the left ventricle. The effects of pericardial pressure have not been considered.

3.11 *The myocardium—ventricular wall stresses*

As well as predicting deformation patterns, the analyses performed for the various ventricular-stiffness configurations also produce stress values in the myocardium. With the general 3-dimensional representation used here there are six stresses, at any one point, to be considered. These are the three direct and three shear stresses. The most easily interpretable of these are the maximum principal stress and the hydrostatic stress.

Figs. 6 and 7 show the maximum principal stresses in the two isotropic models of M and N. Comparison of the hydrostatic stress to the maximum principal stress for each model shows the same

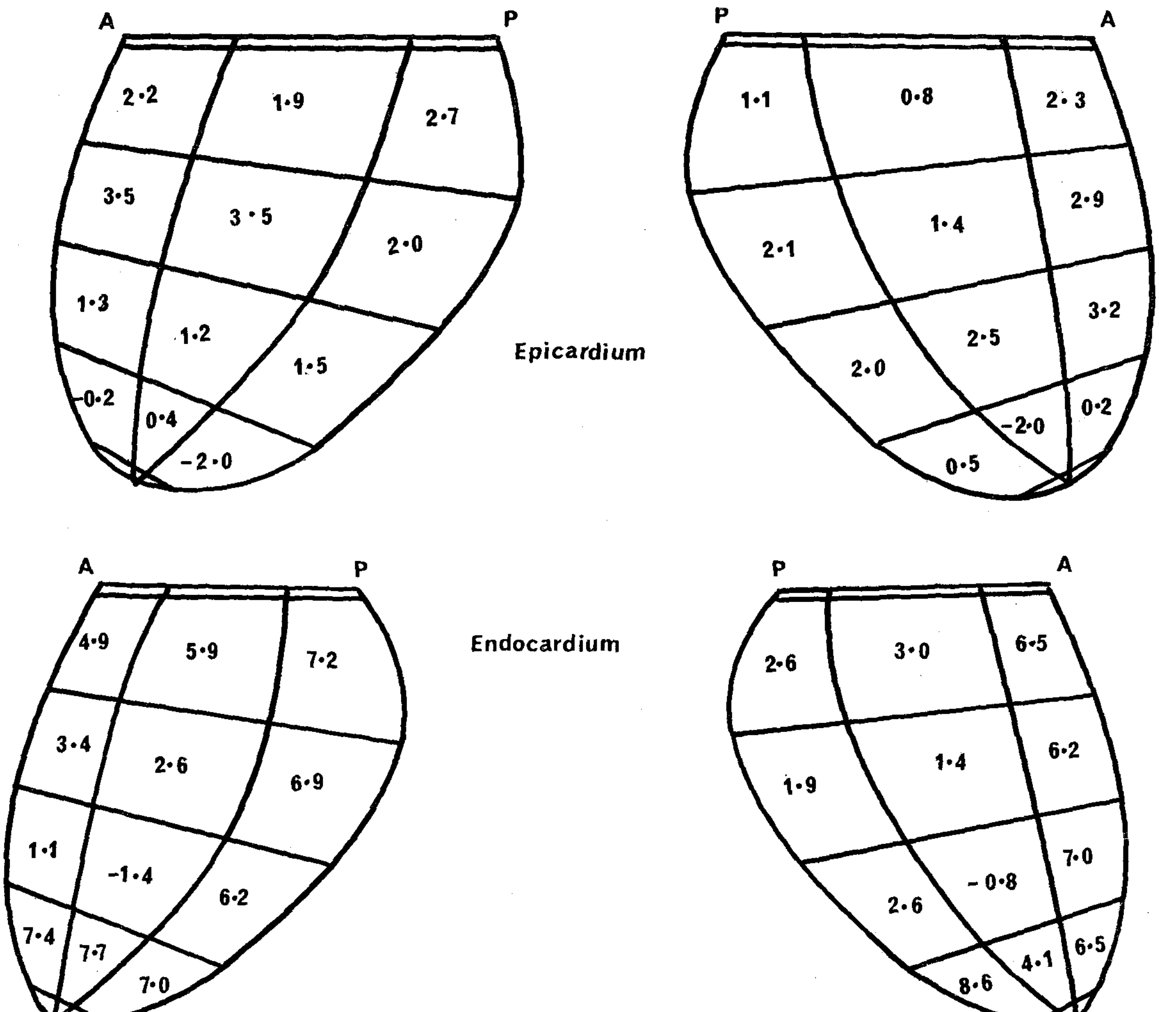

Fig. 8 Patterns of the maximum absolute values of principal direct stresses per unit internal pressure—Patient N

pattern with regard to the rise in stress in a region but the principal values are approximately double.

Normalising the principal stress for model N by dividing by the pressure (Fig. 8) allows these stresses to be compared with those of model M. As shown in the Figures, the maximum principal stress for N is higher on the inner surface and lower on the outer surface compared with M, so that myocardial stress in the two patients was not widely different.

The increase in stress towards the apex reported by HAMID and GHISTA (1974) was found and not the drop suggested by PAO *et al.* (1974) and RITMAN (1975).

From the ventricular models with infarcts, as would be expected, the stress in the flexible infarct drops, while the surrounding myocardial stress rises. For the stiff infarct the reverse applies, with the stress in the infarct rising and the surrounding myocardial stress dropping. Most important, perhaps, is that the stress pattern is no longer seen as being simple, whereby any one cross-section may be used to describe it.

4 Discussion

In the previous Section the results of stress analyses were presented for two ventricular geometries with various patterns of myocardial stiffness, and from these certain biomechanical conclusions could be drawn. If the necessarily limited nature of the study is borne in mind it is possible briefly to consider some of the clinical implications which may arise.

The influence of the stiffness of the annulus fibrosus on the whole pattern of ventricular movement was noted. This might imply that the surgical addition of a rigid prosthetic heart valve might well effectively stiffen the area of the annulus fibrosus to which it will be attached. This may then produce a similar disadvantageous effect of the displacement characteristics of the left ventricle in diastole as that of the modelled collageneous annulus. Such an effect has been noted by BRISTOW (1970) and SUTTON *et al.* (1977).

Consideration of the deformation patterns produced from the cases with various locations and stiffnesses of infarction seem to indicate that an apical infarct may have less influence on the ventricular filling than one at the base. An anterior vertical infarct was almost as disadvantageous as one at the base. These results, in combination with those relating to the annulus fibrosus, therefore suggest that any stiffening at the base of the ventricle is disruptive to normal displacement, while those near the apex have less effect.

The application of the finite-element method in its 3-dimensional form has indicated that the distribution of stiffness influences the passive deformation of the myocardium of the left ventricle in diastole. It is apparent that elastic modulus is not the unique determinant of wall position in diastole as has been generally assumed. As stiffness is dependent both on the geometry of a structure and the constitutive relationship for its material, it is essential that for a specific clinical application both of these physical properties be known with some confidence. When experimental cardiology has developed such that these data can be provided for individual patients then computer modelling will enable correlations between mechanical stress levels and aetiology to be examined. Nevertheless, even with the quantity and quality of data available from current cardiographic techniques, analyses of the type presented here could be applied to a large range of subjects with various forms of cardiac disease. This aspect is discussed by VINSON *et al.* (1979) elsewhere. The results should then make it possible to identify factors which singly or in combination might have clinical importance and also contribute to a better understanding of the mechanics of the left ventricle in both normal and diseased states.

References

BRISTOW, J. D. *et al.* (1970) Systolic and diastolic abnormalities of the l.v. in coronary artery disease: Studies in patients with little or no enlargement of ventricular volume. *Circulation* **42**, 219.

GHISTA, D. N., SANDLER, H. and VAYO, W. H. (1975) Elastic modulus of the human left ventricle. *Med. & Biol. Eng.* **13**, 151–162.

GIBSON, D. G. and BROWN, D. J. (1974) Relation between diastolic left ventricular wall stress and strain in man. *Br. Heart J.* **36**, 1066–1077.

GRATZ, C. M. and BLACKBERG, S. N. (1935) Engineering methods in medical research. *Mech. Eng.* **57**, 217–220.

HAMID, M. S. and GHISTA, D. N. (1974) Finite element analysis of human cardiac structures. *Finite Element Methods in Engineering*, University of NSW, 337–348.

JANZ, R. F. and GRIMM, A. F. (1972) Finite element model for the mechanical behaviour of the left ventricle, *Circulation Res.* **30**, 244–252.

KEITH, A. (1907) An account of the structures concerned in the production of the jugular pulse. *J. Anat. & Physiol.* **42**, 1–25.

LIAN, C. (1909) De la physiologie de l'appareil. Valvulair mitral. *J. Physiol. et Pathol. Géneral*, **11**, 597.

LOWER, R. (1669) *Tractatus de Corde*. London, John Redmayne.

LUNDIN, G. (1944) Mechanical properties of cardiac muscle. *Acta Phys. Scand.* **I**, Suppl. XX.

MIRSKY, I. (1976) Assessment of passive elastic stiffness of cardiac muscle. *Prog. Cardiovasc. Dis.* **18**, 277.

NECKYFAROW, C. W. and PERLMAN, A. B. (1976) Deformation of the human left ventricle: material and geometric effects. 4th New England Bioengineering Conference, 169–172.

NIKRAVESH, E. P. (1976) Optimisation of finite element analyses with special reference to three-dimensional left ventricular dynamics. Doctoral thesis, Tulane University.

PAO, V. C. *et al.* (1974) Finite element analysis of left ventricular myocardial stresses. *J. Biomech.* **7**, 469–477.

PARMLEY, W. W. *et al.* (1973) *In vitro* length-tension relations of human ventricular aneurysons. *Am. J. Cardiol.* **32**, 889–894.

RITMAN, E. L. (1975) Left ventricular function and myocardial contractility. *Proc. Mayo Clin.* **50**, 147–156.

SANDLER, H. and DODGE, H. T. (1963) Left ventricular tension and stress in man. *Circulation Res.* **XIII**, No. 2.

SANTAMORE, W. P. *et al.* (1976) Left ventricular effects on right ventricular developed pressure. *J. Appl. Physiol.* **41**, 925–930.

STREETER, D. D. (1969) A rational model for myocardial stress analysis based on fibre orientation and curvature. Doctoral thesis, Catholic University of America.

SUTTON, M. G. ST. JOHN, TRAILL, T. A., GHAFOUR, S. A., BROWN, D. J. and GIBSON, D. G. (1977) Echocardiographic assessment of l.v. filling after mitral valve surgery. *Br. Heart J.* **39**, 1283.

VINSON, C. A., GIBSON, D. G. and YETTRAM, A. L. (1979) Analysis of left ventricular behaviour in diastole by means of the finite element method. *Br. Heart J.* **41**, 60–67.

YETTRAM, A. L. and VINSON, C. A. (1979*b*) Geometric modelling of the human left ventricle. *J. Biomech. Eng. ASME* (in press).

YETTRAM, A. L. and VINSON, C. A. (1979*a*) Orthotropic elastic moduli for left ventricular mechanical behaviour. *J. Biomed. Eng. & Comp.*, **17**, 25–30.

ZIENKIEWICZ, O. C. (1973) *The finite element method in engineering science.* Wiley, London.

ZIMMERMAN, J. (1966) The functional and surgical anatomy of the heart. *Ann. R. Coll. Surg. Eng.* **39**, 348–366.