Comprehensive model for the simulation of left ventricle mechanics

Part 1 Model description and simulation procedure

M. Perl A. Horowitz S. Sideman

The Julius Silver Institute of Biomedical Engineering, Technion—Israel Institute of Technology, Haifa 32000, Israel

Abstract—*A comprehensive numerical model, baesd on the finite-element method, for the continuous simulation of the complete cardiac cycle is presented. The model uses real, measured* in vivo, *three-dimensional geometry of the ventricle, and accounts for the anisotropy of the ventricular wall, the large deformations it undergoes during the cardiac cycle, the material nonlinearity of the myocardium and its mechanical activation. The simulation process is carried out incrementally while adjusting the mechanical activation for each increment so as to produce the same change in cavity volume as that measured experimentally. A detailed analysis of a complete cycle of the canine heart is presented in Part 2 of the paper.*

Keywords—*Finite-element model, Left ventricle mechanics, Mechanical simulation*

Med. & Biol. Eng. & Comput., 1986, **24**, 145–149

1 Introduction

THE FORMULATION of models for the description of the mechanics of the left ventricle (LV) has been intensively studied. The main parameters involved in the analysis of LV mechanics are:

(*a*) the real irregular geometry of the left venticle
(*b*) the anisotropy of its walls
(*c*) the large deformations the LV undergoes during the cardiac cycle
(*d*) the material nonlinearity of the myocardium
(*e*) the mechanical activation of the heart muscle.

The recently reviewed models (PAO, 1980; PERL and HOROWITZ, 1985; YETTRAM and VINSON, 1979*a*) can be divided into two main groups. The models of the first group account for a large number of the above listed mechanical parameters while compromising with the geometric configuration by describing the LV as a circular cylinder (ARTS *et al.*, 1979; FEIT, 1979; TOZEREN, 1983). The models of the second group emphasise the geometric accuracy, but account for a considerably smaller number of mechanical factors (HEETHAAR *et al.*, 1977; YETTRAM *et al.*, 1983).

The available models usually describe only part of the cardiac cycle, concentrating mostly on the diastolic phase in which the ventricular wall is assumed to be passive. This is due to the evident difficulty in simulating the mechanical activation of the heart muscle, which is especially significant in the systole. However, even when only the diastole is considered, the elastic moduli of the walls must be considerably adjusted (MOSKOWITZ *et al.*, 1978; YETTRAM and VINSON, 1979*b*). Thus, negative elastic moduli values are sometimes required to reproduce the measured diastolic volume/pressure relationships. This result leads to questioning the mechanical meaning of the elastic moduli and the common assumption that the myocardium may be treated throughout the diastole as an entirely passive material.

The present paper attempts to present a comprehensive simulation model, based on the finite element (FE) technique, of the mechanics of the LV, which includes the real geometry of the ventricle as well as the various mechanical parameters formerly mentioned. The model describes the entire cardiac cycle in a continuous manner, bringing the ventricle to its initial state at the end of the simulation process. The simulation procedure is presented, following a detailed discussion of the various assumptions used in the model.

2 Mechanical model

The formulation of the proposed mechanical model of the LV considers the ventricle as a three-dimensional thick-walled vessel which is acted upon by the cavity blood pressure and by the active forces which develop in the muscle fibres. The basic assumptions regarding the LV are:

(*a*) the LV is an irregular three-dimensional body, without rotation axis or symmetry planes (the right ventricle is not included in the model)
(*b*) the amount of deformation the ventricle undergoes during the cardiac cycle implies that a large deformation analysis is needed
(*c*) the myocardium is an elastic nonlinear material
(*d*) the myocardium is an anisotropic material, comprising a fibrous structure embedded in an isotropic bulk material
(*e*) the active forces which develop in the myocardial fibres are accounted for throughout the whole cycle

First received 5th November 1984 and in final form 16th April 1985

(*f*) the endocardium is subjected to a uniform time-varying normal pressure
(*g*) inertial forces are insignificant during the cardiac cycle (i.e. the system is treated quasistatically).

2.1 *LV geometry*

The geometric irregularity of the LV and its material and geometric nonlinearities make the FE method the most suitable technique for solving the problem. The three-dimensional reconstruction of the LV is based on the geometric data provided by RITMAN *et al.* (1980) and obtained by computerised tomography. The original data were reduced to a series of 16–19 parallel transverse cross-sections of the ventricle, spaced 3 mm apart, taken at a frequency of 60 times per second. The endocardial and epicardial contours are delineated on each section by some 50–500 points.

Once the LV shape is determined it is necessary to discretise it prior to the FE analysis. Therefore, based on some of the above points, a three-dimensional FE grid is constructed. An example of the grid is presented in Fig. 1, showing that the FE mesh consists of two types of isotropic elements: the three-dimensional solid ones and the truss elements. The LV is shown to be divided into two layers of three-dimensional elements representing the myocardium, reinforced by truss elements which serve to model the fibrous structure of the muscle.

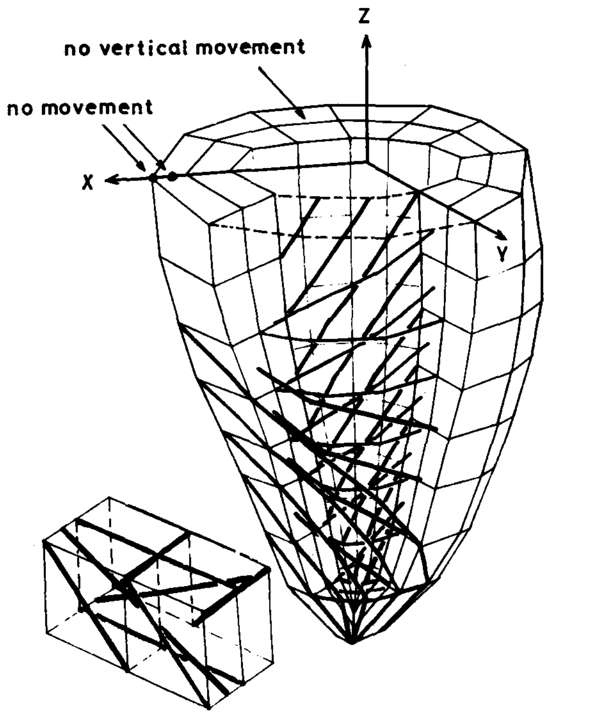


Fig. 1 *Breakdown of the LV into finite elements. The LV wall consists of two layers of three-dimensional elements, combined with five layers of truss elements. The kinematic boundary conditions restricting rigid body displacements and rotations are also indicated*

2.2 *Anisotropy of the LV wall*

The anatomic studies of the fibrous structure of the myocardium (GREENBAUM *et al.*, 1981; STREETER and HANNA, 1973; STREETER, 1979) indicate a gradual fan-like distribution of the fibre directions across the ventricular wall. The common, although probably simplistic, way of describing the fibres' architecture in the ventricle is to assume a linear variation of the fibre angle, from $+60°$ to the circumferential direction at the endocardium to $-60°$ at the epicardium. Alternatively, one can divide the wall into a number of layers, each with a discrete orthotropic direction. The directional anisotropy of the ventricular wall is modelled here by superimposing the truss elements on the mesh of the three-dimensional elements. In the forthcoming implementation the set of truss elements consists of five layers, each with a different fibre orientation which approximately describes the direction of the muscle fibres at the corresponding wall thickness. Thus the fan-like field of the myocardial fibres is lumped into a discrete number of truss layers. By ascribing different cross-sections to the truss elements in each layer it is possible to model the desired experimentally determined (STREETER and HANNA, 1973; STREETER, 1979) quantity of fibres in a particular layer of the ventricular wall. The truss elements have two roles:

(*a*) To increase the stiffness of the wall in the fibre direction relative to the normal direction. This is in accordance with the results of PAO *et al.* (1980) from mechanical experiments on myocardial strips under uniaxial loading. In these experiments the uniaxial stress/strain relationship is given by:

$$\sigma = \frac{c}{k}\left[\exp(k\varepsilon) - 1\right] \tag{1}$$

Eqn. 1 holds for the whole strip as well as for the fibre direction in each layer but with different values for the experimental constants c and k. In the present model the parameters for the truss elements are ascribed the experimentally determined values for the fibre direction (PAO *et al.*, 1980). The values for the three-dimensional elements are calculated so as to obtain for the combination of the two such a behaviour under uniaxial stretch which fits the measured parameter values for the whole strips (HOROWITZ, 1984; PAO *et al.*, 1980).

(*b*) to define the direction of the active forces which develop within the muscle as a result of the shortening of the sacromeres. Since the sacromeres are aligned along the fibres, the direction of the force coincides with the fibre direction.

2.3 *Geometric nonlinearity*

The significant change of shape that the LV undergoes during the cardiac cycle requires a large deformation analysis, and a stepwise approach to the system is adopted. Hence, the structure is loaded incrementally, applying relatively small loading steps and assuming that the resulting incremental displacements are small enough to be considered geometrically linear. The geometry of the structure is updated at each step by a Lagrangian-like scheme. The next step geometry is thus obtained by adding the incremental nodal displacement u to the current $(i-1)$th nodal co-ordinates, i.e.

$$\{X\}_i^N = \{X\}_{i-1}^N + \{u\}_i^N \tag{2}$$

where N is the node number and i is the ith loading increment.

2.4 *Material nonlinearity*

As noted above, the stress/strain relationship for a uniaxial myocardial tissue in the passive state is given by eqn. 1 (CAPPELLO *et al.*, 1981; DEMER and YIN, 1983; KITABATAKE and SUGA, 1978; PAO *et al.*, 1980; PINTO and FUNG, 1973). In the absence of quantitative data for the multi-axial mechanical properties of the muscle, the values of the uniaxial experimental constants (c and k) in eqn. 1, are adopted for the description of the material behaviour of the three-dimensional, as well as for the truss, finite elements. The exponential material law is implemented in the model in a piecewise-linear manner, in which the stress/strain relationship for each of the linear segments is given by

$$(\sigma)_k = \left(\frac{d\sigma}{d\varepsilon}\right)\varepsilon_k \qquad (3)$$

where k denotes the kth linear segment. This approach necessitates the definition of a single parameter for each element to characterise the stress state and consequently to ascribe the appropriate Young's modulus to this element. Whereas the axial stress may serve this purpose for the truss element, it is necessary to express the stress tensor in the three-dimensional elements by one parameter, such as the Von-Mises effective stress:

$$\sigma_{eff} = \{\tfrac{1}{2}[(\sigma_{XX} - \sigma_{YY})^2 + (\sigma_{YY} - \sigma_{ZZ})^2 + (\sigma_{ZZ} - \sigma_{XX})^2] + 3\tau_{XY}^2 + 3\tau_{YZ}^2 + 3\tau_{ZX}^2\}^{1/2} \qquad (4)$$

Note that the Young's modulus for each individual element is updated for each load increment, based on the accumulated effective stress σ_{eff} prevailing in that element.

The second elastic modulus to be ascribed is the Poisson ratio. The common approach regarding soft tissues, including the myocardium, is to consider them as incompressible and thus to ascribe a Poisson ratio of 0·5. However, the experimental data employed in the present study indicate a volumetric strain of approximately 10 per cent for the ventricular wall (see Fig. 11 in part 2, HOROWITZ *et al.*, 1986). Hence, to account for this phenomenon, a lower value of $\nu = 0{\cdot}45$ is chosen for the three-dimensional element material.

After each updating of the Young's moduli, the incremental displacements of the present (i) step, $\{u\}_i$, are computed and consequently the new stress increment $[\Delta\sigma]_i$ is also calculated, using the global stiffness matrix of the previous step, $[K]_{i-1}$:

$$[K]_{i-1}\{u\}_i = \{F\}_i \qquad (5)$$

where $\{F\}_i$ are the nodal forces at the ith step. The calculation of the stiffness matrix for the first step, $[K]_0$, is based on the given LV geometry at QRS and on the initial Young's moduli. For each time step, the stiffness matrix is recalculated based on the updated geometry and the updated Young's moduli. Hence, the current total stress is given by the sum of the previously accumulated stress and the present stress increment:

$$[\sigma]_i = [\sigma]_{i-1} + [\Delta\sigma]_i \qquad (6)$$

2.5 *Active forces in the LV*

The deformation of the wall of the LV and the stress level of this active muscle during the cardiac cycle are strongly affected by the active forces developing within the myocardial fibres, particularly during the systole. The common modelling approach to the diastole assumes that the myocardium behaves as a strictly passive material, deformed by the applied internal cavity pressure. However, if the myocardium is ascribed the experimentally measured passive mechanical properties (PAO *et al.*, 1980), the given increase in the intraventricular pressure is too small, at least for the early part of the diastole, to produce the measured change in the cavity volume. Therefore, part of the deformation of the LV during diastole is probably due to the gradual release of the negative fibre strains built up during systole. This aspect is also reflected in the experimentally reported existence of restoring forces or elastic recoil during the filling phase (SONNENBLICK, 1980).

The activation process is implemented in the present simulation by applying time-varying forces through the truss elements. Whereas the orientation of these forces coincides with that of the truss elements, their magnitude is assumed to have the following form:

$$F_j^{ac} = \alpha(t)A_j \qquad (7)$$

where F_j^{ac} denotes the active force applied to the jth truss element, A_j is the cross-sectional area of the element and $\alpha(t)$ is a time-varying activation factor. The active force is chosen to be proportional to the cross-sectional area of the element, since this area represents the relative number of fibres in any given layer of the LV wall. The mechanical activation is assumed to have a uniform spatial distribution throughout the myocardium, and the same $\alpha(t)$ is ascribed to all the truss elements at any instant of the cardiac cycle.

With the absence of quantitative information regarding the activation function, the activation factor $\alpha(t)$ is taken as the independent variable of the simulation. Its value is chosen so that the numerically simulated pressure/volume loop coincides with that measured *in vivo*. The appropriate calculation steps are:

(*a*) The ventricular volume is maintained constant during the isovolumic stages by equating the volume change generated by the active forces with the opposite volume change caused by the internal pressure.
(*b*) The ventricular contraction in the ejection phase is produced by increasing the active forces and overcoming the opposite distending effect of the internal cavity pressure.
(*c*) The ventricular expansion in the filling phase is obtained by the combined effects of the internal pressure rise and the simultaneous decrease in the active forces.

3 Boundary conditions

To perform the FE analysis it is necessary to prescribe appropriate boundary conditions which constrain the free body movement of the LV, ascribe the appropriate pressure distribution on the endocardium and define the active forces through the truss elements.

The movement of the LV in the thorax is partially restrained by the right ventricle, the atria, the pericardial sack as well as the aorta and the lungs. Obviously, none of these constraints constitutes an absolute clamp, and the quantitative information required to express the anatomical environment of the ventricle in terms of kinematic boundary conditions for the mechanical model is still unavailable. Nevertheless, such boundary conditions are necessary to prevent rigid body displacements and rotations. The appropriate conditions satisfying these requirements while causing only minimal and local distortion of the elastic field under consideration here are (HEETHAAR *et al.*, 1977; YETTRAM *et al.*, 1983):

(*a*) clamping two adjacent nodal points on the LV base
(*b*) prevention of the vertical movement of all the nodal points on the LV base (see Fig. 1).

The blood volume contained by the LV cavity applies a normal pressure and a shearing stress on the endocardial wall. Information describing the distribution and magnitude of the shearing stress is nonexistent, and its inclusion in the model is not yet feasible. The normal cavity pressure acting on the endocardium is obviously a function of time and space, but since the data regarding the local pressure gradients in the LV is very limited, a uniform normal pressure is applied at any given time. The time variance of the pressure is based on experimental measurements acquired simultaneously with the ventricular geometry.

4 Simulation procedure

The entire cardiac cycle is simulated here in a continuous manner. The onset of the electrical activation, the QRS, is chosen to be the starting point of the simulation because at this particular state no active strain exists in the fibres and only a small intramural pressure prevails. In principle, any other arbitrary point could have been chosen as the initial state, bearing in mind that the resulting stresses may deviate from the true value since the initial state might not be stress-free.

The convergence criterion for the simulation process would evidently be the similarity of the calculated shape of the LV to the experimentally measured one at each time step.

To allow the comparison between these two three-dimensional irregular configurations on the one hand, and to provide a feasible algorithm for the process on the other hand, only one geometrical parameter is chosen at this stage to characterise the deformation. In the present work this parameter is chosen to be the cavity volume, since it is a global indicator of the LV deformation and also has a physiological significance. This approach, although not ensuring an exact local match between the calculated and given shapes, provides a general similarity between them.

The simulation runs as follows: given the three-dimensional geometry of the LV at the QRS, an increment of intraventricular pressure is applied as well as a tentative increment of active forces. Then a finite-element analysis is performed, yielding the deformed configuration of the LV due to the above applied loads. The new cavity volume is compared with the experimentally measured one. If the numerical value of the total cavity volume deviates from the experimental finding by more than 1 per cent the present step is repeated with adjusted incremental active forces until it converges. The 1 per cent value was chosen since it is within the accuracy limits of both the experimental measurements and the numerical calculation of the LV volume. Once the above requirement is fulfilled, the geometry of the LV and the stresses are updated according to eqns. 2 and 6, respectively. Then the effective stress in each element is calculated using eqn. 4 and an appropriate Young's modulus is assigned to it. The process is repeated for the whole cardiac cycle and ends when the initial state is reached. A schematic flow chart of the process is given in Fig. 2.

The guiding principle in choosing the magnitude of each pressure increment is to justify the geometric and material linearity approximation within that increment. Nevertheless, the total number of increments should be such as to make the whole process computationally feasible.

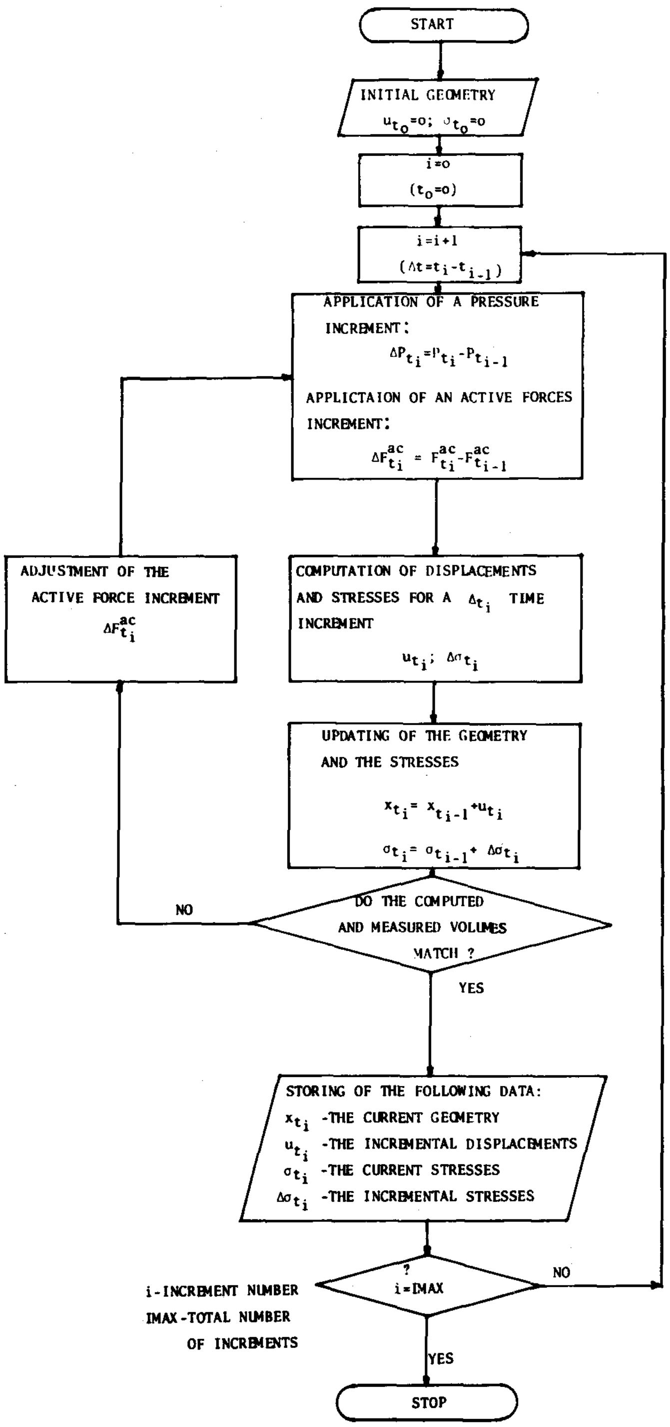


Fig. 2 *Flow chart of the simulation process indicating the method of applying pressure and active forces steps at each increment and of fitting the calculated and given cavity volumes*

5 Concluding remarks

The proposed model combines a large number of properties characterising the mechanical behaviour of the left ventricle: its real irregular shape, the anisotropy of the ventricular wall, its geometric and material nonlinearities and the activation of the myocardium. In so doing the present model takes into consideration more factors than other existing models, and carries out a continuous simulation of the entire cardiac cycle.

Acknowledgment—This study was supported by the Michael-Kennedy-Leigh Foundation, London and sponsored by the MEP Groups, Women's Division, American Technion Society, USA. Our thanks to Professor Y. C. Pao, University of Nebraska, Lincoln, Nebraska, for his encouragement and advice throughout this study.

References

ARTS, T., RENEMAN, R. S. and VEENSTRA, P. C. (1979) A model of the mechanics of the left ventricle. *Ann. Biomed. Eng.*, **7**, 299–318.

CAPELLO, A., COMINCIOLI, V., MINELLI, R., REGGIONI, C. and RICCIARDI, L. (1981) Study and parameters identification of a rheological model for excised quiescent cardiac muscle, *J. Biomech.*, **14**, 1–11.

DEMER, L. L. and YIN, F. C. P. (1983) Passive biaxial mechanical properties of isolated canine myocardium. *J. Physiol.*, **339**, 615–630.

FEIT, T. S. (1979) Diastolic pressure-volume relation and distribution of pressure and fiber extension across the wall of a model left ventricle. *Ann. Biomed. Eng.*, **28**, 143–166.

GREENBAUM, R. A., HO, S. V., GIBSON, D. G., BECKER, A. and ANDERSON, R. H. (1981) Left ventricular fiber architecture in man. *Br. Heart J.*, **45**, 248–263.

HEETHAAR, R. M., PAO, Y. C. and RITMAN, E. L. (1977) Computer aspects of three-dimensional finite element analysis of stresses and strains in the intact heart. *Comput. Biomed. Res.*, **10**, 271–285.

HOROWITZ, A. (1984) A numerical model for the simulation of the mechanics of the left ventricle. M.Sc. thesis, Technion-Israel Institute of Technology, Haifa, Israel.

HOROWITZ, A., PERL, M., RITMAN, E. and SIDEMAN, S. (1986) A comprehensive model for the simulation of left ventricle mechanics. Part 2. Implementation and results analysis. *Med. & Biol. Eng. & Comput.*, **24**, 150–156.

KITABATAKE, A. and SUGA, H. (1978) Diastolic stress-strain relation of nonexcised blood perfused canine papillary muscle. *Am. J. Physiol.*, **234**, 416–419.

MOSKOWITZ, S. E., LEWIS, B. S., HALON, D. A., AMAR, R. and GOTSMAN, M. S. (1978) Computer prediction of left ventricular compliance throughout diastole in normal patients. *Eur. J. Cardiol.*, 7/Suppl., 121–132.

PAO, Y. C. (1980) Discussion: geometric modelling of the human left ventricle. *J. Biomech. Eng.*, **102**, 274–275.

PAO, Y. C., NAGENDRA, G. K., PADIYAR, R. and RITMAN, E. L. (1980) Derivation of myocardial fiber stiffness equation based on theory of laminated composites. *Ibid.*, **102**, 252–257.

PERL, M. and HOROWITZ, A. (1985) Limitations of finite element models of the left ventricle—review and future prospects, Proc. Henry Goldberg Workshop Simul. Imag. Cardiac Syst., SIDEM, S., and BEYAR, R. (Eds.), Martinus Nijhoff.

PINTO, J. G. and FUNG, Y. C. (1973) Mechanical properties of the heart muscle in the passive state. *J. Biomech.*, **6**, 597–616.

RITMAN, E. L., KINSEY, J. H., ROBB, R. A., GILBERT, B. K., HARRIS, L. D. and WOOD, E. H. (1980) Three-dimensional imaging of heart, lung and circulation. *Science*, **210**, 273–280.

SONNENBLICK, E. H. (1980) The structural basis and importance of restoring forces and elastic recoil for the filling of the heart. *Eur. Heart J.*, **1** Suppl., 107–110.

STREETER, D. D. and HANNA, W. T. (1973) Engineering mechanics for successive states in canine left ventricular myocardium. *Circ. Res.*, **33**, 656–664.

STREETER, D. D. (1979) Gross morphology and fiber geometry of the heart. In *Handbook of physiology, vol. 1 The heart.* BERNE, R. M., SPERELAKIS, N. and GEIGER, S. R. (Eds.), Am. Phys. Soc., Bethesda, Section 2: The cardiovascular system, 61–112.

TOZEREN, A. (1983) Static analysis of the left ventricle. *J. Biomech. Eng.*, **105**, 39–46.

YETTRAM, A. L. and VINSON, C. A. (1979*a*) Geometric modelling of the human left ventricle. *Ibid.*, **101**, 221–223.

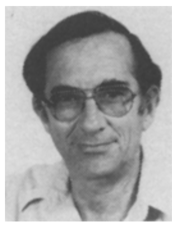
YETTRAM, A. L. and VINSON, C. A. (1979*b*) Orthotropic elastic moduli for left ventricular mechanical behaviour. *Med. & Biol. Eng. & Comput.*, **17**, 25–30.

YETTRAM, A. L., VINSON, C. A. and GIBSON, D. G. (1983) Effect of myocardial fiber architecture on the behavior of the human left ventricle in diastole. *J. Biomed. Eng.*, **5**, 321–328.

Authors' biographies

Mordechai Perl is Senior Lecturer in the Department of Mechanical Engineering of the Technion—Israel Institute of Technology (IIT). He received the B.Sc., M.Sc. and D.Sc. in Aeronautical Engineering from the Technion in 1968, 1970 and 1979. His current research interest in biomedical engineering is cardiac mechanics and his interests in mechanical engineering are fracture mechanics, dynamic fracture and fatigue and creep crack growth.

Arie Horowitz is an instructor in the Department of Mechanical Engineering at the Technion, IIT. He received the B.Sc. degree in Architecture and the M.Sc. degree in Bio-Medical Engineering in 1980 and 1984, both from the Technion, IIT. He is currently studying for the D.Sc. degree at the same institute. His research interests are cardiac mechanics and application of the finite-element method to tissue mechanics.



Samuel Sideman is Director of the Julius Silver Institute of Biomedical Engineering and Head of the Department of Biomedical Engineering at the Technion, IIT. He is a member of the editorial board of a number of scientific journals and a recipient of national and international awards, including the Roy J. Matas Winnipeg Professorship in Biomedical Engineering. He received the B.Sc. Chem. Eng. from the Technion, in 1953, the M.Ch.E. from the Polytechnic Institute of Brooklyn, New York, USA, in 1955 and the D.Sc. from the Technion in 1960. He has been with the Technion faculty since graduation. His teaching and research interests are transport phenomena and modelling and simulation of physiological systems.