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OPTIMIZATION OF LEFT VENTRICULAR FIBRE ORIENTATION OF THE NORMAL HEART FOR HOMOGENEOUS SARCOMERE LENGTH DURING EJECTION

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ABSTRACT

During the ejection phase of the cardiac cycle, left ventricular muscle fibres shorten while generating force. It was hypothesized that fibres are oriented in the wall such that the amount of shortening is the same for all fibres. We evaluated this hypothesis for the equatorial region of the left ventricle.

In a finite element model of left ventricular wall mechanics fibre orientation was quantified by a helix angle which varied linearly from the inner to the outer wall. Fibre length was characterized by sarcomere length, set at 1.95 μm everywhere in the passive state of 0 transmural pressure. For a cavity pressure of 15 kPa, considered representative for ejection, inhomogeneity in mechanical loading was expressed by the variance of the sarcomere length. The variance was minimized by adapting the transmural course of fibre angle. First, only the slope was optimized and in a second optimization this was done for both slope and intercept.

Optimal helix fibre angles were 69.6° endocardially, 0° at the middle of the wall and -69.6° epicardially for the first optimization and 78.2°, 20.7° and, -36.7° respectively for the second. Sarcomere length changed from 1.95 to 1.975 \pm 0.012 and 1.981 \pm 0.004 μm (mean \pm SD) respectively. Conclusion: After optimization calculated helix fibre angles were in the physiological range. Describing the transmural course of fibre angle with slope and intercept significantly improved homogeneity in mechanical load.

KEYWORDS: finite element modelling – helix – fiber angle

INTRODUCTION

During the cardiac ejection phase the muscle fibres in the left ventricular wall shorten while loaded by mechanical stresses. An earlier model study (Bovendeerd et al., 1992) indicated that the distribution of mechanical load in the wall depends strongly on the local orientation of muscle fibres. In the model by Bovendeerd et al. (1992) the shape of the wall was prolate spheroidal. The mechanical behaviour

of cardiac tissue was nonlinear, anisotropic and elastic. Contractile force generated by fibres depended on time, sarcomere length and sarcomere velocity of shortening. Fibre orientation was quantified by the helix fibre angle (Fig. 1). Small changes in fibre orientation had large consequences for the distribution of fibre stress and strain in the wall during the ejection period. However, the pressure-volume relationship of the left ventricular cavity was nearly unaffected by changes in fibre orientation.

In the normal heart mechanical load is likely to be distributed evenly over all myocardial structures (van der Vusse et al., 1990). In a computer simulation (Arts et al., 1994) it was postulated that the summed actions of many regional adaptations to mechanical

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load evens mechanical load. Local wall mass was assumed to increase with increasing mechanical load. Fibre orientation was assumed to change locally so that fibre shortening during the ejection phase was the same everywhere. Using these assumptions in a cylindrical model of left ventricular wall mechanics, several features of cardiac anatomy formed automatically: for instance the helical arrangement of fibres in the wall, the control of filling volume of the cavity, and the control of wall mass.

In another study (Peskin, 1989) left ventricular fibre architecture was derived assuming axial symmetry and that the wall is thin compared to other ventricular dimensions. Fibre bundles were also assumed to have a constant cross-sectional area along their length. They could generate force only in the fibre direction. Fibres were embedded in a fluid-like material sustaining only a hydrostatic pressure. Assuming the ventricle to be in mechanical equilibrium a fibre architecture was calculated which was found to be physiological. However, fibres did not necessarily experience the same tensions, so that mechanical load was not necessarily the same for all fibres.

In the present study we investigated whether fibres could be arranged in the left ventricular wall such that they all experience the same mechanical load. No detailed hypotheses were formed concern-

ing the mechanism of this minimization. Our model differs from that of Peskin (1989) in that the wall thickness need not be small and also in that cardiac tissue is composed of fibres embedded in an elastic material instead of a fluid. Secondly, if such a minimum was found then the resulting fibre architecture was compared with the measured anatomical fibre structure of the heart.

SETUP OF THE NUMERICAL MODEL

Theoretical background

In Fig. 2 the pressure-volume relation of the left ventricle during a cardiac cycle is shown schematically. During diastole, the relation follows a path determined by the passive material properties of the myocardium. According to the variable elastance concept (Sagawa, 1978), the ventricle stiffens gradually until maximum elastance has been reached at the end of the ejection phase. We aim to find a cardiac structure so that during the ejection phase, shortening of the myocardial fibres is as uniform as possible. Simulation of a whole cardiac cycle with a finite element model of ventricular mechanics is very time consuming. To find an optimum, many simulations have to be performed. We have applied a simplification to save a major amount of computing time.

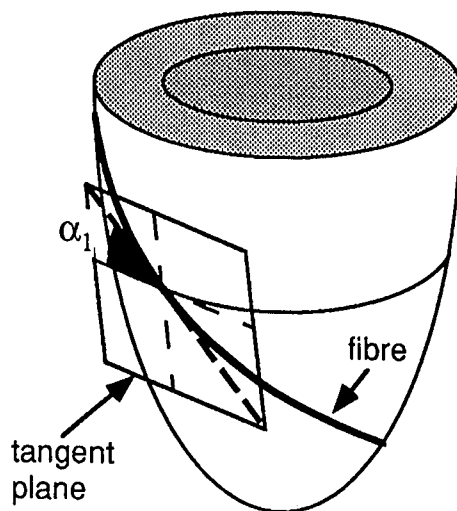


Fig. 1. THE HELIX FIBRE ANGLE, α_1 , IS DEFINED AS THE ANGLE BETWEEN THE LOCAL CIRCUMFERENTIAL DIRECTION AND THE PROJECTION OF THE FIBRE PATH ON THE PLANE PERPENDICULAR TO THE WALL. Helix fibre angle is negative at the indicated site.

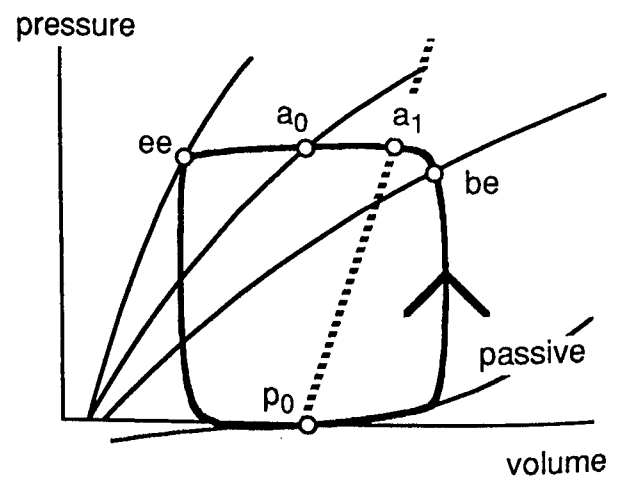


Fig. 2. PRESSURE-VOLUME RELATION OF THE LEFT VENTRICULAR CAVITY DURING A CARDIAC CYCLE. During diastole the relation follows a curve, as determined by the passive elastic properties of the myocardium. During systole, the pressure-volume relation varies as a function of time, reaching the highest level of mechanical activation at the end of systole.

First, the requirement of a uniform distribution of sarcomere shortening from the beginning (be) to the end (ee) of ejection has been transformed to the requirement of a uniform distribution of the change in sarcomere length for a given change in left ventricular volume ($\Delta l_s/\Delta V_{lv}$) during the ejection phase. In Fig. 2 this interval is marked by the states a_0 and a_1 . It has been reported that in diastole sarcomere length is almost uniform ($1.95 \pm 0.03 \mu\text{m}$) (Grimm et al., 1980). We assumed that at zero diastolic pressure, sarcomere length is $1.95 \mu\text{m}$ while stress is zero everywhere. Then, keeping volume constant, the material has been activated or stiffened to reach the state a_0 , while no deformation took place. Then in the activated state volume is increased to the state a_1 . In the optimization procedure, the ventricular structure should be adapted so that the change in sarcomere length from state a_0 to a_1 was made as uniform as possible. In a further simplification we moved directly from state p_0 to state a_1 by increasing activation in direct proportion to the volume. This problem has been converted to stiffening of the sarcomere stress-strain relation, so that the pressure volume path p_0 to a_1 is followed. Thus, the problem has been reduced to finding a structure so that the change in sarcomere length is uniform by pressurizing the cavity with (unphysiologically) stiff myocardial fibres.

Geometry

In the model, the reference state of the left ventricle was defined as the passive state with zero transmural pressure and a uniform sarcomere length of $1.95 \mu\text{m}$. The endo- and epicardial wall surfaces between the equator and apex were represented by prolate spheroid-like shapes. We assumed point-symmetry with respect to the centre of the equatorial plane. Hence, only the mechanics of the region between the equator and apex is modelled. In the left ventricles of potassium-arrested dog hearts, measured cavity volume is $40 \pm 9 \text{ ml}$ and wall mass is $145 \pm 19 \text{ g}$ (McCulloch et al., 1989). Using a myocardial tissue density of 1.05 g/cm^3 , a cavity to wall volume ratio of 0.3 can be calculated. Assuming that the base-to-equator height is twice that of the apex-to-equator height, the volumes between equator and apex of the wall (including the papillary muscle volume of 4 ml) and of the cavity were calculated as 86 and 25 ml respectively. The equatorial wall thickness was twice that of the apex. The long to short axis ratio of the midwall surface was 2.08.

Fibre orientation

Fibre orientation was quantified by the helix fibre angle (Fig. 1). The transmural course of the helix fibre angle in the left ventricular wall in the reference state was described by a straight line. Fractional wall thickness was described by a coordinate which takes on the values -1 , 0 , $+1$ at the inner, mid-, and outer walls respectively. A slope and an intercept were used to describe the linear relationship between the helix fibre angle and the fractional wall thickness coordinate.

Material properties

In the model, myocardial tissue was assumed to consist of stiff muscle fibres embedded in a soft matrix material. Each constituent contributed to the total stress in the tissue. The constitutive behaviour of the soft matrix was assumed to be elastic and transversely isotropic (Demer & Yin, 1983; Yin et al., 1987), the stiffness in the fibre direction being twice as high as that in a cross-fibre direction. The stresses resulting from deformation of the passive tissue increased exponentially with the strains (Bovendeerd et al., 1992). In the real left ventricle the contractile force generated by a fibre is a function of time, sarcomere length, and shortening velocity of sarcomeres (ter Keurs et al., 1988; ter Keurs et al., 1980). The mean shortening velocity of a sarcomere during ejection may be estimated from Rodriguez et al. (1992) as $1.5 \mu\text{m s}^{-1}$. At this shortening velocity a muscle fibre is able to generate about 50 % of its maximum contractile force (Daniels et al., 1984). To take sarcomere velocity of shortening during ejection into account in the model, we have adopted the variable elastance concept (Sagawa, 1978) causing the contractile force that a muscle fibre can generate during ejection to be about halved. As discussed above the fibres generate no force at $1.95 \mu\text{m}$. At sarcomere lengths only slightly greater (about $2.0 \mu\text{m}$) the simulated contractile force was a good approximation of the experimentally observed force. The ventricular wall was further assumed to be incompressible and homogeneous in its material properties.

Calculation procedure

Mechanical load was calculated from equations expressing conservation of momentum in the wall. Inertial (Moskowitz, 1981; Peskin, 1989) and gravitational effects are neglected in the model. Hence conservation of momentum expresses the static equi-

librium of forces in the wall due to blood pressure in the cavity and internal stresses in the wall. In principle the solution of the equations can be expressed in terms of the displacements of all points in the wall. In this study the equations were solved using the finite element (FE) method, in which the displacement of a finite number of points, so-called nodes, in the wall is calculated. In between those points the displacement field was interpolated quadratically. The

nodes were grouped into 20-node elements which can be regarded as building blocks of myocardial tissue.

The endocardial surface was loaded normally by cavity pressure while the epicardial surface experienced no external load. The loading of the left ventricular wall was rotationally symmetric. Since there are no inhomogeneities in material properties the ventricle maintains its rotational symmetry in the

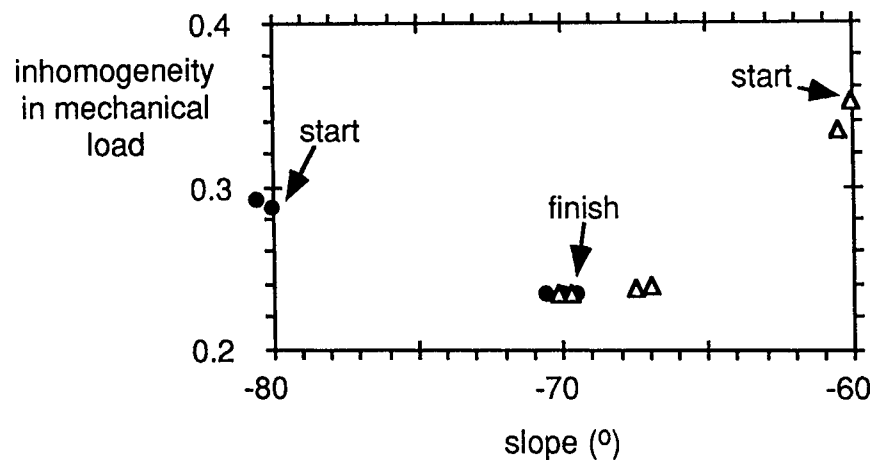


Fig. 3. OPTIMIZATION OF SLOPE OF LINEAR TRANSMURAL COURSE OF HELIX FIBRE ANGLE. INHOMOGENEITY IN MECHANICAL LOAD IS DEFINED AS THE VARIANCE OF THE MOMENTANEOUS SARCOMERE LENGTH AT MID-EJECTION NORMALIZED TO MEAN SARCOMERE SHORTENING. Symbols denote finite element evaluation: (●) starting at a slope of -80° , and (Δ) starting at -60° .

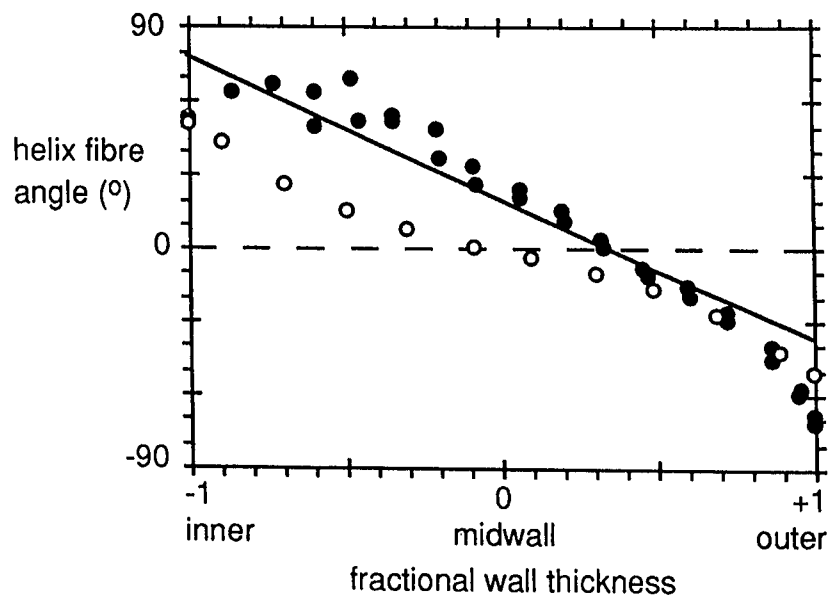


Fig. 4. OPTIMAL TRANSMURAL COURSE OF HELIX FIBRE ANGLE (CONTINUOUS LINE) FOR SECOND OPTIMIZATION. Also shown are measurements from: (○) Streeter *et al.*, (1969), and (●) Streeter *et al.*, (1979).

loaded state. To save computing time only a quarter section of the ventricle was considered. The presence of the remaining three-quarter sections and of the region between equator and base was accounted for by kinematic boundary conditions.

The left ventricle was inflated from the reference state, in which cavity pressure is 0 kPa gauge, to 15 kPa. In the loaded state sarcomere lengths were calculated at 729 points in the wall.

Quantification of mechanical load

During the ejection period all fibres generate contractile force. The mechanical state of the ventricle is then largely governed by the fibres. This state was therefore deemed more likely to determine fibre orientation than a diastolic loading state. No sensors of mechanical stress are known at a cellular level. Mechanical load was hence quantified in terms of strain. Sarcomeres are the basic contractile protein units of all cardiac fibres. The strain occurring in sarcomeres is a good measure of fibre strain. Mechanical load may hence be conveniently expressed in terms of sarcomere strain. Furthermore, computer model studies (Bovendeerd, 1990) have shown that the spatial distribution of fibre strain does not change much during ejection. Knowledge of the fibre strain distribution at one point during ejection gives a good approximation of the distribution over the whole ejection period. Because sarcomeres all have the same length of $1.95 \mu\text{m}$ in the reference state p_0 , and by assumption at a_0 (Fig. 2), fibre strain with respect to this state can also be expressed in terms of sarcomere length. Inhomogeneity in mechanical load was defined as the variance of the calculated sarcomere length at mid-ejection normalized to the average amount of sarcomere shortening. 324 Points in the equatorial region of the wall were used to determine the variance.

Optimization of fibre orientation for homogeneous sarcomere length in loaded state

Load inhomogeneity was assumed to depend only on the fibre orientation parameters. The relation is implicit and the FE model is required to evaluate it for a given fibre orientation. The inhomogeneity was minimized by adapting the fibre orientation in an automated procedure. Such a minimization process entails repeated evaluation of load inhomogeneity for various input fibre orientations. To reduce the computational effort we resorted to the concept of

sequential approximate optimization (Haftka & Gürdal, 1992). This is useful when evaluation of the function to be minimized, the so-called objective function, is computationally expensive. Briefly, based on a few exact evaluations of the objective function for very similar input parameters, the objective function is replaced by an approximation function whose dependence on the input parameters is known explicitly. The approximation function is then minimized using standard techniques. At the proposed minimum the process of generating a new approximation function may be repeated until a particular convergence criterion is satisfied.

Optimizations performed

In the first optimization, the transmural variation of the helix fibre angle was described by a straight line whose slope was to be optimized. In the middle of the wall the helix fibre angle was set to 0° . Load inhomogeneity was minimized by systematically changing the slope of the line. Two simulations were carried out with different starting points; in one a slope of -60° was used and in another -80° was the starting point. In a second optimization, load inhomogeneity was minimized by optimizing both the slope and the intercept of the linear transmural helix fibre angle distribution.

RESULTS

The result for the optimization of the slope of the linear relation between transmural position ($-1 \leq \bar{\xi} \leq +1$) and the helix fibre angle, α_1 , is shown in Fig. 3. The solution was independent of the slope with which the simulation was started. Within 3 iterations a fibre orientation that minimized mechanical load was found. The optimal slope was -69.6° ; helix fibre angle was 69.6° endocardially and -69.6° epicardially. With the optimal slope, sarcomere length in the equatorial region changed from 1.95 in the reference state to $1.975 \pm 0.012 \mu\text{m}$ (mean \pm standard deviation) in the loaded state. The root mean square (RMS) errors of the calculated fibre orientation were 16.5° and 22.3° with regard to measurements by Streeter et al. (Streeter et al., 1969) and Streeter et al. (Streeter, 1979) respectively.

In the second optimization both the slope and the intercept of the linear transmural course of helix fibre angle were optimized. Starting with a slope of -

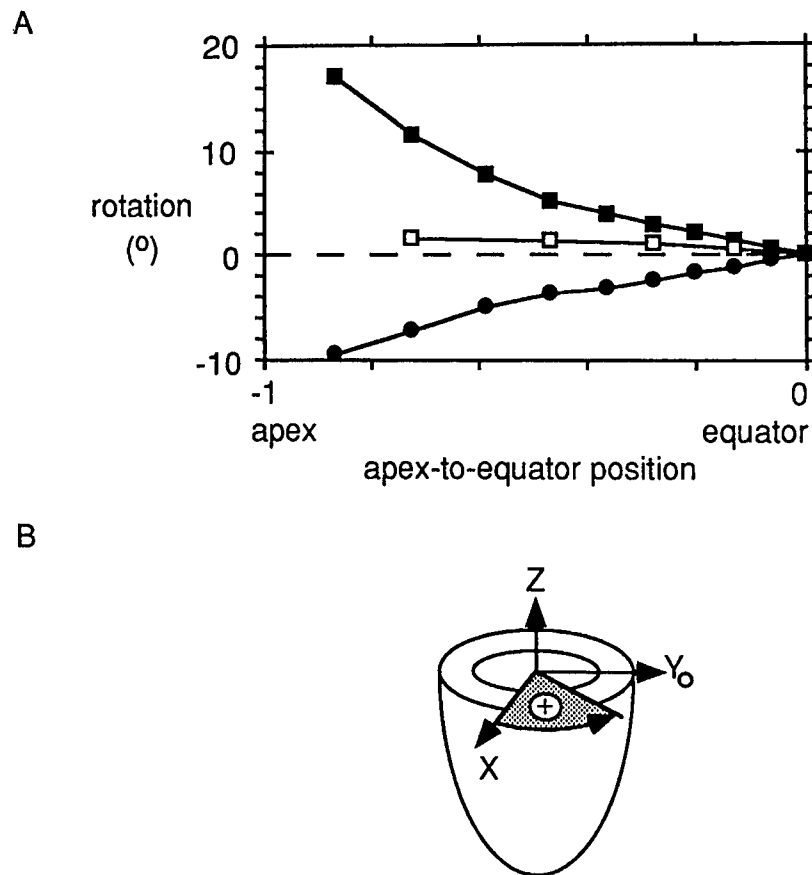


Fig. 5. PANEL A: CALCULATED ROTATION ANGLES FOR OPTIMAL SLOPE OF HELIX FIBRE ANGLE (FIRST OPTIMIZATION). Rotations are calculated at a cavity pressure of 15 kPa relative to the zero pressure state. Near the apex the transmural differences in rotation angle are large. (■) inner wall, (□) midwall, (●) outer wall. Panel B: Sign convention for rotation angles.

70° and an intercept of 15°, an optimum was found in 3 iterations. The optimal slope and intercept were -57.5° and 20.7° respectively. With the optimal fibre orientation sarcomere lengths in the equatorial region changed from 1.95 μm in the reference state to $1.981 \pm 0.004 \mu\text{m}$ in the loaded state. The calculated optimal fibre orientation compared well with measurements of Streeter et al. (Fig. 4). The RMS errors of the calculated fibre orientation were 23.4° and 14.1° with regard to measurements by Streeter et al. (1969) and Streeter et al. (1979) respectively.

In the loaded state rotations about the long ventricular axis of the inner with respect to the outer wall were large for the first optimization with transmural differences up to 26° (Fig. 5). In the second optimization transmural differences in rotation were no greater than about 18°.

DISCUSSION

A computer model was developed to simulate the regional and global mechanics of the left ventricular wall. The distribution of mechanical load in the left ventricular wall could be calculated for a given wall geometry, fibre direction, material properties and left ventricular loading pressure. Load inhomogeneity was quantified as the variance of the sarcomere length in the neighbourhood of the equator at a cavity pressure of 15 kPa. Fibre orientation was adapted so that load inhomogeneity was minimal. Adapting the transmural course of helix fibre angle using an intercept and a slope allowed significant improvement in the attainable degree of homogeneity in mechanical load.

The use of more parameters to describe the transmural course of fibre orientation may further improve load homogeneity. Bovendeerd et al. (1992) used 4 parameters to describe the transmural course

helix fibre angle. Fibre orientation was optimized manually to make the transmural course of fibre stress during ejection in the equatorial region as homogeneous as possible. Mean fibre stress during ejection was more homogeneous if near the inner and outer walls the helix fibre angle distribution was nonlinear. Measurements (Fig. 4) made by Streeter and colleagues (Streeter et al., 1969; Streeter, 1979) also indicate nonlinear behaviour near the walls. Furthermore, there is some evidence that helix fibre angle increases slightly in the base-to-apex direction (Streeter et al., 1969; Streeter & Hannah, 1973).

Cross-over of fibres from the inner to the outer walls has been observed, most clearly in the apical and basal regions (Streeter et al., 1975). Including this fibre orientation parameter in our model may further improve load homogeneity. Fibre cross-over may considerably reduce transmural differences in rotation about the long axis of the ventricle (Bovendeerd et al., 1994). Maximum transmural differences in rotation in the performed simulations range from 18° to 26° near the apex (Fig. 5). The rotation of the endo- and epicardial surfaces in normal humans has been measured using a magnetic resonance imaging technique (Buchalter et al., 1990). The transmural differences in rotation angles during systole were at most 8° near the apex. It is conceivable that fibre cross-over should affect transmural differences in rotation. Without cross-over, fibres lie in concentric shells. Forces tending to cause transmural differences in rotation then cause large deformations before balancing internal forces are generated because the passive matrix from which they arise is soft. Fibre cross-over will ensure that counteracting forces are generated at smaller deformations.

It has been assumed that sarcomeres in the passive ventricular wall all have the same length when the transmural pressure in the cavity is 0. There is some evidence that in this state epicardial sarcomeres are about $0.1 \mu\text{m}$ longer than those near the endocardial wall (Rodriguez et al., 1993). For simplicity these differences have not been incorporated in the model.

To be sure that the obtained optimum is the true global optimum the optimization should be carried out more than once with different starting values. In the first optimization the optimum seems to be independent of the starting point. The second optimization was not performed from different starting points. Hence we cannot be sure that the absolute optimum has been found.

The optimizations carried out consider the whole cardiac structure at once; a global measure of load inhomogeneity was defined and minimized. To the best of our knowledge no structures in the heart are capable of collecting all the information that we have used to optimize with. It is more likely that optimization occurs in a spatially more restricted environment such as that of the cell and a close neighbourhood around it (Arts et al., 1994).

In conclusion, a linear transmural course of helix fibre angle has been calculated which makes mechanical load near the equator as homogeneous as possible. Optimizing both the slope and the intercept of the transmural distribution of the helix fibre angle together makes mechanical load more homogeneous than if only the slope were optimized. The calculated optimal transmural course of helix fibre angle compared well with measurements, the RMS error of the most optimal calculated helix fibre orientation with respect to measurements by Streeter et al. (1979) being 14.1° .

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