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Comparison of Young Laplace law and finite element based calculation of ventricular wall stress: Implications for post infarct and surgical ventricular remodeling

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Abstract

Objectives—Both the Young Laplace law and finite element (FE) based methods have been used to calculate left ventricular (LV) wall stress. We tested the hypothesis that the Young Laplace law is able to reproduce results obtained with FE method.

Methods—Magnetic resonance (MRI) images with non-invasive tags were used to calculate 3D myocardial strain in five sheep 16 weeks after anteroapical myocardial infarction and in one of those sheep 6 weeks after a Dor procedure. Animal specific FE models were created from the remaining five animals using MRI images obtained at early diastolic filling. FE based stress in the fiber, cross fiber and circumferential directions was calculated and compared to stress calculated with (Young Laplace law) and without (Modified Laplace) the assumption that wall thickness is very much less than the radius of curvature.

Results—First, circumferential stress calculated with the Modified Laplace law is closer to results obtained with the FE method than stress calculated with the Young Laplace law. However, there are pronounced regional differences with the largest difference between Modified Laplace and FE occurring especially in the inner and outer layers of the infarct borderzone. Also, stress calculated with Modified Laplace is very different than stress in the fiber and cross fiber direction calculated with FE. As a consequence, the Modified Laplace law is inaccurate when used to calculate the effect of the Dor procedure on regional ventricular stress.

Conclusion—The FE method is necessary to determine stress in the LV with post infarct and surgical ventricular remodeling.

Keywords

Myocardial Infarction; Strain; Stress; Finite Element; Laplace

II. INTRODUCTION

Regional coronary blood flow, [1] myocardial oxygen consumption, hypertrophy, [2] and remodeling are all determined by ventricular wall stress. In addition, a reduction in wall stress has been used to justify a number of new cardiac operations including the Batista operation [3], the Acorn cardiac support device [4] and the Myocor Myosplint. [5] However, Huisman previously showed that it is impossible to directly measure regional in vivo myocardial stress because of tethering from surrounding myocardium. [6,7] As a consequence, ventricular wall stress is usually calculated with force balance methods such as the Young Laplace law. [8,9] Unfortunately, localized shape change in and around the myocardial infarction (MI) and regional changes in systolic and diastolic material properties [10] make the left ventricle (LV) a mechanically complex structure. The accuracy of force balance based calculation of LV stress may be, therefore, severely limited.

An alternative approach to quantifying ventricular wall stress and stiffness is mathematical modeling based on the conservation laws of continuum mechanics, the most versatile of which is the finite element (FE) method. [11] The FE method used in this study for continuum analysis of the heart includes several features that are uncommon in conventional FE methods. First, the constitutive relationship is non-linear and anisotropic [12,13] with direction based directly on measured 3-D myofiber angle distributions. [14,15] Next, the models undergo large or finite deformation with difference in dimensions between the end-diastole (ED) and end-systole (ES) that is greater than 10%. More recent models now also include the transmural heterogeneity of cellular excitation-contraction coupling mechanisms. [16]

We calculated stress with (Young Laplace law) and without (Modified Laplace) the assumption that wall thickness is very much less than the radius of curvature with large deformation finite element methods using data collected from sheep after antero-apical myocardial infarction (MI) and after Dor procedure. We tested the hypothesis that stress in the circumferential direction calculated with Laplace law is equal to values obtained with the FE method. Since stress in the cross fiber direction is likely to cause volume overload hypertrophy and stress in the fiber direction is likely to cause pressure overload hypertrophy, we also compared stress calculated using force balance with the magnitude of stress in the fiber and cross fiber directions calculated with the FE method.

II. MATERIAL AND METHODS

Animals used in this study were treated in compliance with the “Guide for the Care and Use of Laboratory Animals” prepared by the Institute of Laboratory Animal Resources, National Research Council, and published by the National Academy Press, revised 1996. Magnetic resonance (MRI) images with non-invasive tags used to calculate 3D myocardial strain in five sheep 16 weeks after anteroapical myocardial infarction and in one of those sheep 6 weeks after Dor procedure.[17] These experimental results were previously reported. [17,18]

A. Force balance calculations

i. The Young Laplace law—The endocardial contour was divided into 32 points. As seen in Figure 2, 3 adjacent points on the endocardial contour were identified and used to determine the radius of curvature. [21] Two line segments were created by connecting point 1 and point 2, point 2 and point 3, where two bisectors to these two line segments were then constructed. The radius of curvature for point 2 was calculated from the distance from point 2 to the bisector’s intersection. Note that short axis contours near the apex (infarct area) were not used because the image plane in that area cuts the LV wall obliquely.

The derivation of the Young-Laplace law is as follows:

$$\sigma [\pi (r+h)^2] = P\pi r^2 \quad (1)$$

where P is intracavitary pressure, r is the endocardial radius of curvature and h is the wall thickness. Equation 1 simplifies to:

$$\sigma = \frac{Pr}{h(2 + \frac{h}{r})} \quad (2)$$

if $h \ll r$ Equation 2 is reduced to the Young Laplace law:

$$\sigma = \frac{Pr}{2h} \quad (3)$$

ii. Modified Laplace law—Throughout the paper we refer to Equation 3 as ‘Young Laplace’ and Equation 2 as ‘Modified Laplace’.

B. FE Model

Analysis of borderzone function using FE models has been previously reported. [22]

i. Finite element method—Animal specific FE models were created from MRI images at the early diastole, which was considered as the initial unloaded reference state.[23]. Representative surface and finite element meshes are seen in Figure 3. Myofiber angles of -37° , 23° and 83° were assigned at the epicardium, midwall and endocardium, respectively, in the remote and borderzone regions [24]. At the aneurysm region, fiber angles were set to 0° [25]. Circumferential displacement of basal epicardial nodes was constrained.

ii. Loading—The inner endocardium wall was loaded to the measured *in-vivo* ED and ES LV pressures.

iii. Constitutive relationship—Passive [13] and active myocardial [26] constitutive relationships have been previously described. The active myocardial material property law was implemented using a user-defined material subroutine in LS-DYNA (Livermore Software Technology Corporation, Livermore, CA). Diastolic [27] and systolic [26] material parameters have been previously reported.

iv. Material property optimization—The method of myocardial material property optimization was previously described by Sun et al. [29] Specifically, the commercial FE optimization software, LS-OPT [30], was used to find the optimal value of T_{max} for each region.

Simulations were performed on a small Linux cluster (7 nodes; each node with two AMD Opteron 240 processors and 1 gB memory).

C. Statistical analysis

All values are expressed as mean $\pm$ standard deviation (SD) and compared by repeated measures analysis using a mixed model to test for both fixed and random effects (Systat Software, Inc., Chicago, IL). The statistical model was as follows:

$$s = \text{Method} + \text{Region} + \text{Layer} \quad (5)$$

where Method, Region and Layer are dummy variables and where Method is either Laplace or FE, Region is either BZ or remote and Layer is either Endo, MID or Epi. The statistical significance of individual group comparisons was tested using the Student *t* test with the Bonferroni correction for multiple comparisons. Significance was set at *p* less than 0.05.

III. RESULTS

Computational time was recorded while performing the FE method on a single FE model. Time for FE simulation was 385 seconds.

A. Average circumferential stress

Table 1 shows average LV wall stress in the circumferential direction calculated using the Young Laplace law, the Modified Laplace law and the finite element method. The average wall thickness to radius of curvature ratio was 0.22 ± 0.022 at end diastole and 0.24 ± 0.030 at end systole. As a consequence, circumferential stress calculated with the Young Laplace law and the Modified Laplace law is significantly different.

Furthermore, circumferential stress calculated with the Modified Laplace is closer to results obtained with the FE method than stress calculated with the Young Laplace law. Specifically, average Modified Laplace stress was 26% higher than FE based calculation of circumferential stress at end-diastole ($p=0.006$) but was not different at end-systole.

B. Regional variation

There were significant regional differences. Stress calculated with the Modified Laplace was higher than circumferential stress calculated with the FE method in the remote (37%, $p=0.048$) but not borderzone regions (Figure 4A) and higher than circumferential stress calculated with the FE method in the endocardial layer (93%, $p<0.001$) at end-diastole (Figure 4B).

Furthermore, stress calculated with the Modified Laplace law was higher than circumferential stress calculated with the FE method in the borderzone (Modified Laplace: 52%, $p=0.048$) but not in the remote myocardium (Figure 5A) and higher than circumferential stress calculated with the FE method in the endocardial layer (93%, $p<0.001$) at end-systole (Figure 5B).

C. Comparison with fiber and cross fiber stress calculated with the FE method

The magnitude of stress calculated with the Modified Laplace law was very different than stress in the fiber and cross fiber direction. Average fiber stress with the FE method was 3.119 ± 1.784 kPa at end-diastole and 38.618 ± 10.653 kPa at end-systole. Average cross fiber stress with the FE method was 1.935 ± 1.112 kPa at end-diastole and 8.800 ± 7.102 kPa at end-systole. Average Modified Laplace stress was 22% higher than FE based calculation of cross fiber stress end-diastole ($p<0.001$) and 121% higher at end-systole ($p<0.001$).

Once again, there were significant regional differences. Stress calculated with the Modified Laplace law was higher than cross fiber stress calculated with the FE method in the remote region (22%, $p=0.012$) at end-diastole (Figure 4A). Stress calculated with both the Modified Laplace law was substantially lower than fiber stress calculated with the FE method in both

remote (Modified Laplace: 64%, $p<0.001$) and borderzone regions (Modified Laplace: 35%, $p=0.012$) at end systole (Figure 5A). Finally, Stress calculated with the Modified Laplace law was lower than fiber stress at all layers ($p<0.001$) at end-systole (Figure 5B).

D. Change in stress after Dor procedure

Stress at end systole calculated with the Young Laplace law, Modified Laplace law and the finite element in the infarct borderzone before and after Dor procedure is seen in Table 2. The reduction in average borderzone stress calculated with the Young Laplace and Modified Laplace law was -23.5 and -26.2% respectively. However, the change in regional stress calculated with the finite element method was quite different with a substantial increase in stress in the inner layer (Δ in circumferential stress $+196.7\%$; Δ in fiber stress $+15\%$) and a more pronounced decrease in the outer layer (Δ in circumferential stress -70.4% ; Δ in fiber stress -49.7%).

IV. Comments

The principal findings of our study are 1) circumferential stress calculated with the Modified Laplace is closer to results obtained with the FE method than stress calculated with the Young Laplace law, 2) there are pronounced regional differences between stress calculated with the Modified Laplace law and the finite element method and 3) the magnitude of stress calculated with the Modified Laplace law is very different than stress in the fiber and cross fiber directions calculated with the finite element method.

A. The Finite Element Method as the ‘Gold Standard’

The decision to use the large deformation finite element method as the reference or gold standard is reasonable given that the finite element models were optimized using measured myocardial strain in addition to end-diastolic and end-systolic volumes. A comparison of Laplace’s law with the finite element method in a physical model constructed from materials with known material properties, such as silicone [31] or silicone with embedded elastic fibers, and that has simplified geometry would be interesting and we propose to carry out this study in the future. However, even when completed, physical models of that sort will lack the ability to simulate active contraction.

B. Limitations of the Young Laplace law

There are a number of inaccuracies and limitations associated with the application of Laplace type calculations of left ventricular wall stress [32]. An inaccuracy specific to the Young Laplace law is the restriction that the wall thickness (h) be very much less than the radius of curvature (r). However, the h/r ratio in this study was 0.22 ± 0.022 at end diastole and 0.24 ± 0.030 at end systole. This suggests that Equation 2 (Modified Laplace) which does not have the $h \ll r$ restriction is more reasonable. In fact, we found in all cases that stress calculated with Equation 2 was closer to finite element based calculation of circumferential stress than stress calculated with the Young Laplace law.

C. Stress variation across the LV wall

When a thick-walled pressure vessel is inflated, most of the deformation in the circumferential and longitudinal directions occurs at the inner surface and decreases monotonically towards the outer surface. Thus, we would expect circumferential and longitudinal stress to vary transmurally in a similar manner although the actual transmural variation in these stress components will depend on the relationship between stress and strain (constitutive relation) of the myocardium. For instance, Guccione et al previously measured fiber stress in the LV of a normal dog using a finite deformation finite element

method similar to that used in the current study. [33], Guccione found the transmural gradient in fiber stress was 3.3 kPa at the base and 4.6 kPa at the apex at end diastole. [33] The transmural gradients were small near the LV base during systole were as high as 43 kPa between the mid ventricle and apex. [33] In the present study, transmural gradients in fiber stress were smaller but still significant with a gradient of 1.55 kPa in the remote myocardium at end diastole and 7.23 kPa at end systole. The Young Laplace law is based on a global force balance, which ignores myocardial material properties. Thus, the Young Laplace law can be used to estimate only average stress across the full wall thickness in the circumferential and longitudinal directions.

D. The importance of fiber and cross fiber stress

Another limitation of the Young Laplace and Modified Laplace laws is that they cannot be used to calculate stress in the local muscle fiber or cross fiber direction since stress in the fiber and cross fiber directions are probably the causes of hypertrophy. There is increasing evidence that stress in the cross fiber direction causes eccentric or volume overload type hypertrophy. [34,35] Although evidence is less clear, it is probable that end systolic fiber stress causes concentric or pressure overload type hypertrophy.

E. Implications for the Dor procedure

Regional differences between stress calculated the Young Laplace law and finite element methods occurs especially in the inner and outer layers of the infarct BZ. When coupled with inability of Laplace type methods to measure fiber and cross fiber stress, this leads the Young Laplace and Modified Laplace laws to miscalculate the change in stress in the infarct BZ after Dor procedure. Since the rationale for the Dor procedure in specific and surgical ventricular remodeling in general is reduction in wall stress [36], accurate knowledge of stress in the fiber and cross fiber directions is of obvious importance. Furthermore, in our opinion, improvement in BZ contractility is the primary therapeutic target of the Dor procedure. [22]

F. Utility of the finite deformation type finite element model of the left ventricle

The FE method can be used to simulate the effect of surgical ventriculoplasty [37] and passive constraint on the LV. For instance, Dang and colleagues previously used a finite element model of the finite deformation type to calculate the effect of surgical remodeling on stroke volume and mean fiber stress. [37] Force balance methods such as the Young Laplace law can only calculate stress from an existing LV image obtained during animal experiments or from patients. Thus, it cannot be used as a predictive theoretical tool. In our opinion, this is the greatest limitation of the Young Laplace law.

V. CONCLUSIONS

Circumferential stress calculated with the Modified Laplace law was superior to the Young Laplace law although stress calculated with the Modified Laplace law remained statistically different (higher) than finite element based calculation of circumferential stress at end-diastole.

On the other hand, the magnitude of stress calculated with the Modified Laplace law is very different than stress in the fiber and cross fiber directions calculated with the finite element method. Also, there are pronounced regional differences with the largest differences between Modified Laplace and finite element methods occurring in the inner and outer layers of the infarct BZ. As a consequence, both the Modified Laplace law is inaccurate when used to calculate the effect of the Dor procedure on regional ventricular stress. The

finite element method is necessary to determine stress in the LV with post infarct and surgical ventricular remodeling.

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Figure 1.

A. Long axis MR images from sheep with anteroapical MI. **B.** Short axis MR image. Both images were obtained 16 weeks after MI. Both images have non invasive tissue tags and both images were obtained at end-systole. MI=dykinetic infarct; SI=septal infarct. Note that Panel A has been previously reported. [17]

Figure 2.
Method of measuring endocardial radius of curvature. r is the radius of curvature and h is the wall thickness.

Figure 3. Finite element model of the LV with anteroapical MI. Animal specific contours were generated from MRI (A). Solid mesh (B) was broken into four regions (green = remote, brown and red = borderzone).

Figure 4B.

Stress by Layer at End-diastole

Figure 4.

Stress calculated with Laplace and finite element methods at end-diastole by region (A) and layer of the LV wall (B). Note that both the Young Laplace's law and Modified Laplace law overestimate stress in the circumferential and cross fiber directions in the remote myocardium at end-diastole.

Figure 5A.

Figure 5B.

Figure 5.

Stress calculated with Laplace and finite element methods at end-systole by region (A) and layer of the LV wall (B). Note that the Young Laplace law underestimates stress in the fiber direction at end-systole. The Young Laplace is not different than circumferential stress in the remote myocardium but fails to account for transmural variation in stress. Statistically significant comparisons with cross fiber stress are not marked. See the text for those details.

Table 1

Average stress in the circumferential direction at end diastole and end systole. Units are kPa. Values are mean $\pm$ standard deviation.

	End-diastole	End-systole
Young Laplace	2.69 ± 0.95^a	22.38 ± 7.97^a
Modified Laplace	$2.35 \pm 0.85^{a,b}$	19.43 ± 7.93^a
Finite element	1.87 ± 1.25^b	15.00 ± 10.27

^a is $p < 0.05$ between Young Laplace and Modified Laplace

^b is $p < 0.05$ between Modified Laplace and Finite element.

Stress at end systole calculated with the Young Laplace law, Modified Laplace law and the finite element in the infarct borderzone before and after Dor procedure. Data is from a single sheep. FE is finite element. Note that the change in regional circumferential and fiber stress calculated with the finite element method was quite different especially in the inner and outer layers of the borderzone.

Table 2

	Inner	Midwall	Outer	Average
Before Dor (16 weeks after MI)				
FE (Fiber)	28.39	26.60	26.76	27.25
FE (Cross Fiber)	6.43	9.16	11.82	9.14
FE (Circumferential)	3.63	17.72	14.40	11.91
Young LaPlace				19.71
Modified LaPlace				17.57
After Dor				
FE (Fiber)	32.65	20.22	13.45	22.10
FE (Cross Fiber)	8.33	7.30	6.39	7.34
FE (Circumferential)	10.77	12.66	8.45	10.63
Young LaPlace				15.08
Modified LaPlace				12.96