IDENTIFICATION OF HYPERELASTIC PROPERTIES OF LUMBAR MULTIFIDUS

[1]Terry K Koo and [2]Yongping Zheng

[1]New York Chiropractic College, Seneca Falls, NY, USA
[2]Hong Kong Polytechnic University, Hong Kong, PRC
email: tkoo@nycc.edu, web: http://www.ultrasonic-solutions.com

INTRODUCTION

Quantitative measurements of elastic properties of lumbar multifidus (LM) may have significant values on low back pain (LBP) diagnosis and prognosis. Attempts have been made to quantify paraspinal soft tissue compliance (deformation per unit load) using instrumented indentation [1]. However, these measures depend not only on the elastic properties of the soft tissues, but also on the shape, thickness, and boundary conditions. Therefore, direct comparison among subjects and sites are difficult. We developed a tissue ultrasound palpation system (TUPS) to make in-vivo elasticity measurements of LM [2,3]. Our previous effort focused on small strains and assumed a linear elastic behavior. The current work extends our previous effort by presenting large strain indentation data on LM to test the feasibility of using TUPS and finite element (FE) modeling to determine non-linear elastic parameters of LM through optimization.

METHODS

Experimental Protocol: Since LM is the only muscle that lies on top of the mamillary process at the lumbar region [4], we located and tested 10 mamillary processes in 3 asymptomatic subjects and 1 right-side affected LBP subject. Subjects were instructed to lie down in a prone position on a treatment table. Each indentation trial involved a ramped load on a marked site to ~25% deformation using the TUPS probe at a rate of ~8%/s. During the indentation trials, each subject was instructed to exhale to a comfortable extent and to suspend breathing. Load-deformation data of 5-7 indentation trials were collected at each site, and curve-fitted with a 3rd order polynomial. A probe alignment arm was used to guide the translation of the TUPS probe during indentation.

Test Apparatus: The TUPS probe has a 5 MHz, 11 mm flat-ended cylindrical ultrasound transducer at its tip mounted in series with a 10 N strain gauge load cell. As an operator manually loads the probe on the tissue surface, the ultrasound module of the TUPS continuously emits ultrasound pulses into the soft tissue. At the interface between the LM and the mamillary process, there is normally an abrupt change of the acoustic impedance, thus producing significant reflections for the incident ultrasonic wave. Hence, thickness and deformation of the LM can be determined from the timing of the echo signals using a cross-correlation procedure, whereas the applied force is measured by the load cell.

Finite element Modeling: The indentation experiment was modeled as a two-dimensional, finite-deformation, axisymmetric problem using Abaqus. The TUPS probe was modeled as a rigid body with a frictionless contact. A vertical displacement that corresponds to 25% tissue deformation was prescribed to the FE model and the corresponding reaction forces were calculated throughout the indentation. The lower surface of the LM was constrained in the vertical direction to account for the fact that the LM lies on top of the mamillary process and assumed a frictionless boundary condition. All soft tissues between the probe and the mamillary process were denoted as LM and modeled as a homogenous layer of hyperelastic material using an order-one Ogden material model. The LM was assumed to be incompressible to simulate the initial apparent incompressibility condition of the soft tissue to indentation. Geometrical nonlinearity was selected within the Abaqus to account for finite-deformation.

Parameter Identification: Two parameters (i.e. an initial shear modulus (μ_1) and a non-dimensional coefficient (α_1)) are required to specify the nonlinear elastic behavior of the LM. Specifically, μ_1 accounts for the linear elastic response of the LM at a small strain and α_1 accounts for the nonlinearity at a larger strain.

A number of steps were carried out to determine the hyperelastic parameters. The technique starts with indentation trials that acquire the force-deformation responses of the LM using TUPS. These responses were used in a least square minimization algorithm to determine the hyperelastic parameters within the FE model that results in a force-deformation response similar to its measured counterpart. FE meshes were automatically generated using the tissue thickness as an input parameter. Initial guess of μ_1 was based on the analytical solution proposed by Hayes [5] and initial guess of α_1 was arbitrarily set to 15. A Matlab program was developed to carry out an iterative adjustment of material parameters, according to Nelder-Mead simplex method. For each iteration, the Matlab program updates the material parameters and links with the Abaqus to invoke the FE analysis. A Python script extracts the simulation results from the Abaqus to the Matlab after each iteration. This iterative process is repeated until termination criteria meet.

An additional parameter was derived to account for the contribution of all hyperelastic parameters to the LM's overall non-linear elastic behavior. Using the hyperelastic parameters computed at each site, we computed simulated force-deformation (F-D) response of a 40mm thick, hypothetical axisymmetric block induced by an 11 mm flat-ended cylinder and calculated the area under the F-D curve (A_{FD}) from 0 to 10 mm deformation using numerical integration. This approach eliminated the F-D variability due to geometry.

RESULTS AND DISCUSSION

The bilateral force-deformation data at L4 and L5 mamillary processes of the right-side affected LBP subject are plotted in Fig. 1. The small difference among the trials suggests that the TUPS measurements are repeatable.

Our FE optimization approach succeeded in fitting the experimental data accurately for all measurement sites. Table 1 summarizes the Ogden parameter set at each site that produced the best fits. The corresponding A_{FD} is also reported. Symmetry ratios (SR) between the left and right A_{FD} were calculated at each level. Initial shear modulus was also calculated using the linear elastic model proposed by Hayes [5]. It appears that quantification of LM stiffness using linear elastic parameters or models can be misleading. A_{FD} appears to be a more useful parameter for quantifying LM stiffness as it accounts for the overall elastic behavior over a large strain range. Our preliminary results also demonstrated that the LBP subject's LM muscles are not necessary stiffer than the asymptomatic subjects. Hence, the clinical impression of palpable rigidity of LM in subjects with LBP is questionable. However, we noted consistency between clinical symptom and elasticity measurement, that is the right side of the right-affected LBP subject is stiffer than the left side.

Future work will incorporate the FE optimization approach on more LBP and asymptomatic subjects at all lumbar levels to explore the potential use of large strain elasticity measurements on LBP diagnosis and/or prognosis.

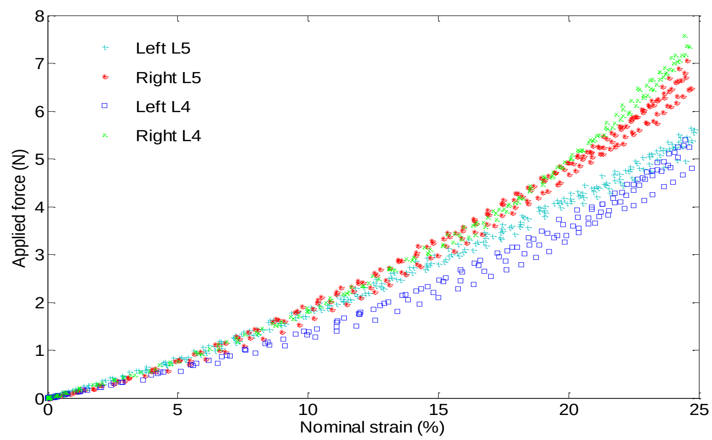

Figure 1: Measured force-deformation data of a right-affected LBP subject at the mamillary processes of the left and right L4 and L5. Data of 5-7 indentation trials are superimposed at each site.

Table 1: Summary of material parameters

Subject#	Site*	Hayes μ_1 (kPa)	Odgen μ_1 (kPa)	Odgen α_1	A_{FD} (Nmm)	SR+
A	L-L4	9.32	10.04	15.77	37.09	
B	L-L3	11.89	10.95	12.95	34.17	
C	L-L4	8.11	8.62	15.81	30.72	0.989
C	R-L4	8.60	9.09	14.93	31.05	
C	L-L5	8.05	8.28	16.36	30.32	0.928
C	R-L5	8.85	8.38	14.58	28.13	
D	L-L4	6.84	7.30	11.15	21.14	0.797
D	R-L4	7.67	8.67	12.48	26.51	
D	L-L5	7.66	9.29	5.730	22.45	0.869
D	R-L5	8.38	9.24	10.25	25.84	

#A, B, C are asymptomatic subjects and D is a right-side affected LBP subject.
*L: left; R: right; +SR: Symmetry ratio between the left and right A_{FD} (≤ 1).

REFERENCES

1. Fischer AA. *Arch Phys Med Rehabil* **68**, 122-125, 1987.
2. Koo TK, et al. *ACC-RAC 2010*, Las Vegas, NV, USA, 2010.
3. Zheng YP, et al. *IEEE Trans Biomed Eng* 43, 912-918, 1996.
4. Macintosh JE, et al. *Spine* **12**, 658-668, 1987.
5. Hayes WC, et al. *J Biomech* **5**, 541-551, 1972.