J Shoulder Elbow Surg (2015) ■, 1-7

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Journal of Shoulder and Elbow Surgery

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A 3D finite element model for geometrical and mechanical comparison of different supraspinatus repair techniques

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Background: Contact pressure and contact area are among the most important mechanical factors studied to predict the effectiveness of a rotator cuff repair. The suture configurations can strongly affect these factors but are rarely correlated with each other. For example, there is a significant difference between the single-row technique and the transosseous or transosseous-like approaches in terms of footprint contact area coverage. A finite element model–based approach is presented and applied to account for various parameters (eg, suture pretension, geometry of the repair, effect of the sutures, geometry of the lesion) and to compare the efficacy of different repair techniques in covering the original footprint.
Methods: The model allows us to evaluate the effect of parameters such as suture configuration and position and suture pretension. The validity of such an approach was assessed in comparing 3 different repair techniques: single row, transosseous equivalent, and double row.
Results: Results from the application of the models show that the double-row and transosseous-equivalent techniques lead to progressive increase of the contact area compared with the single-row approach, supporting the conclusion that transosseous-equivalent fixation leads to an increase of the contact area and a better distribution of the pressure coverage.
Conclusion: The 3-dimensional finite element model approach allows multiple variables to be assessed singularly, weighing the specific influence. Moreover, the approach presented in this study could be a valid tool to predict and to reproduce different configurations, identifying how to reduce the stress over the tendon and when a repair could be effective or not.
Level of evidence: Basic Science Study, Computer Modeling.

Keywords: Shoulder; cuff repair; finite element model; transosseous

The arthroscopic approach in rotator cuff repair is a common surgical procedure. Even if rotator cuff repair studies are widely represented in the literature, the best method able to guarantee a superior functional outcome is still under discussion. Arthroscopic implants are subjected to continuous improvement, permitting complex constructs,

Institutional Review Board approval was not required for this Basic Science study.

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1058-2746/$ - see front matter © 2015 Journal of Shoulder and Elbow Surgery Board of Trustees.
http://dx.doi.org/10.1016/j.jse.2015.09.002

doubling of the anchor rows, or a mixture between transosseous and anchor fixing. The challenge is to create a construct able to apply a higher compression in the footprint area and to maximize the contact extension.[30,32]

The biomechanical superiority of the double-row techniques is well supported in the literature.[20,27] The most frequently inspected factors are the ultimate load to failure in a static setup and the gap formation in a cyclic test.[3,17]

The importance of having a wider and more stable tissue-bone contact during the early phase of tissue regeneration is a key concept presented by many authors.[10,22] The common conclusion is that the methods that produce a smaller contact area and a smaller contact pressure have a potential risk for higher rates of structural failure.

It is important to distinguish between optimized pressure and not maximized because it is now evident that excessive pressure could be deleterious, producing vascular alteration, local stress spikes on the tendon side, and ischemic reaction, and so the optimal amount is that to prevent relative sliding at the bone-tissue interface.

Over the years, many investigations have been conducted of the various reparative approaches with the aim of finding the most effective one. Today, we can identify some key aspects on the basis of the successful rotator cuff repair: initial stiffness and strength of the repair (ultimate tensile strength), gap formation resistance, sliding stability in intra and extra rotation in the immediate postoperative period, maximization of the original footprint coverage, and optimization of the contact pressure at the tendon-bone interface.[1,9,13,15]

Previous works have presented attempts to reproduce the tendon-bone interface with the aim of identifying the most stressed area of the supraspinatus and finding a correlation with tears. Inoue et al[14] found the maximal tensile stress on the articular side of the anterior edge at 90° abduction.

The same results were also confirmed by Wakabayashi et al[33] in 2 findings: first, the articular side is a stress notch; and second, distal shift of stress concentration occurs with the arm in abduction.

Funakoshi et al[12] estimated the suture effect by dividing the experimental measured pressure by the projected suture area. They demonstrated that the stress concentration in a transosseous approach is 23.7% lower than in double-row techniques, without considering the effect of the weaker bone-suture interface. Whereas their findings were not obvious, their approach highlights the importance of the suture effect to the repair.

Another method can be found in Sano et al,[27] who applied a 2-dimensional model to assess the local stress peak due to the presence and position of the defect (lesion). Their study interestingly proved that it is possible to assess a partial intratendinous tear (delamination phenomenon) using a composite material.

Although for different purposes, Sano et al[27] assessed the local peak stress in the bone area close to the anchor insertion (as a function of anchor angle insertion). They focused on the importance of reducing the stress peak with the chosen repair approach not only in the soft tissue but also on the bone side.

Table I Material properties used in the finite element model

Tissue	Young modulus (MPa)	Poisson ratio
Supraspinatus	168	0.497
Cancellous bone	13,800	0.300
Cortical bone	13,800	0.300

The aim of this paper was to present a new approach for the comparison between various repairs in terms of footprint and contact area coverage. The new approach uses a finite element model–based method for evaluating the effect of parameters such as suture configuration and position and suture pretension. Second, the validity of such an approach was assessed in comparing 3 different repair techniques.

Materials and methods

Three-dimensional (3D) finite element models have been conceived to reproduce the various repair techniques: single row (SR), double row (DR), and transosseous equivalent (TE).

A commercial software, ANSYS R14 (ANSYS Inc, Canonsburg, PA, USA), was used as a preprocessor and postprocessor for the finite element analysis. The 3D model was obtained from a computed tomography scan. Computed tomography scans were performed on a cadaver specimen using 1-mm axial slices, a slice increment of <0.625 mm, and a field of view covering the entire humerus and scapula (as indicated by Levy et al[16]).

An anatomic coordinate system was created to measure the orientation in space of the humerus based on anatomic landmarks. The glenoid center point was determined by selecting the smooth surface of the glenoid face and calculating its center. A plane was fit to the selected glenoid face surface to create the glenoid face plane. A neutral inclination axis was defined between the glenoid center point and the trigonum spinae. Inclination was thus measured with respect to the neutral axis, and version was measured with respect to the scapular plane.

Cortical and cancellous bones have been treated as isotropic homogeneous and uniform materials (see Table I for the adopted material properties). The geometrical reproduction of the supraspinatus was based on what was measured in a cadaveric study by Pauly et al.[26]

In our analysis, we used 2 different solid elements, SOLID185 and SOLID285. The number of nodes on average is close to 45,000.

In this work, 3 different repair approaches have been simulated: SR, DR, and TE.[24,25] The abduction angle of the glenohumeral joint was fixed at the initial stage of validation of the model at 0° (position based on the coordinate system as described before). The models used to simulate the various repairs are described in Figure 1 (in all cases, the inserted devices have been

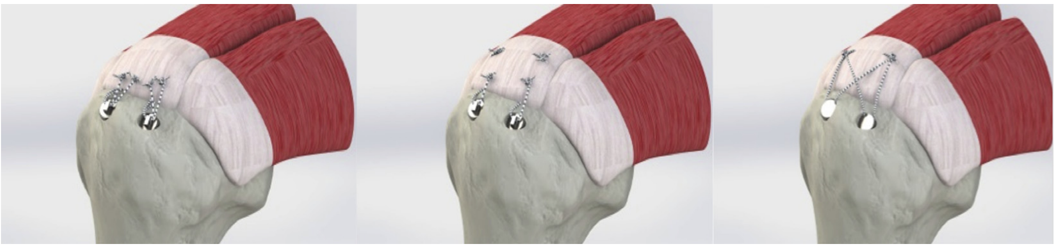

Figure 1 Solid models used in the finite element analysis. From *left* to *right*, single row (SR), double row (DR), and transosseous equivalent (TE).

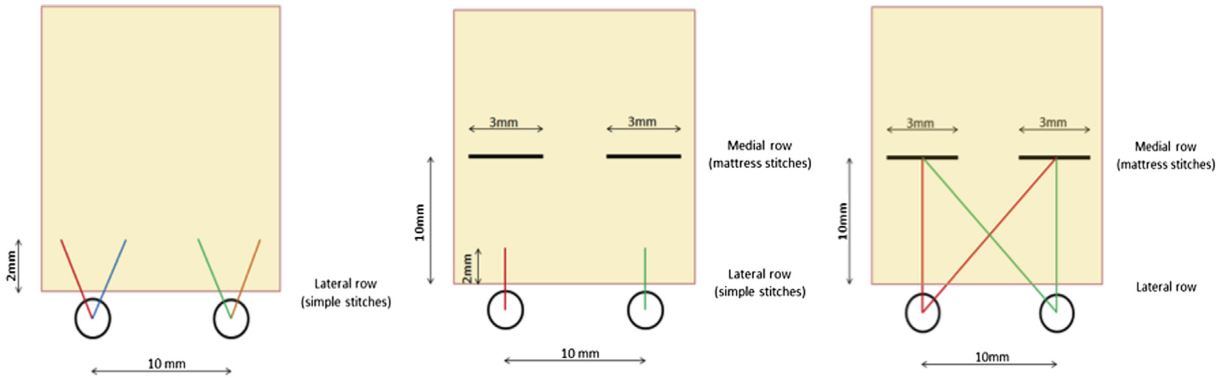

Figure 2 Sketch of devices and suture positions used in the simulation. From *left* to *right*, single row (SR), double row (DR), and transosseous equivalent (TE).

considered not deformable and have been stabilized in the model with a bonded contact). The supraspinatus has been completely detached from the humeral head as is usually done in biomechanical studies, and contact with the humeral head is induced by the knot's pretension of the repair.[12]

The suture 3D models have been introduced in the cadaveric model, and the pretension effect has been simulated by connecting these to various springs having the same stiffness obtained from tensile tests on real sutures. The experimental data, in agreement with what has been reported in the literature,[4] was fixed at 5 N/mm.

The initial geometrical configuration (in terms of supraspinatus positioning) was the same for all the approaches. The geometry of the repair is sketched in Figure 2 for each type of repair.

A representation of the 3D mesh is shown in Figure 3; as visible from the mesh, there is a refinement that interests particularly the tendon and cortical bone in the contact area. Five elements in the thickness direction have been used to capture the gradient in the supraspinatus, and a properly sized mesh has been adopted at the interface with the sutures to capture the mutual pressure transfer.

Sutures were modeled as flexible cylinders having an external diameter of 0.4 mm and a modulus of ultrahigh-molecular-weight polyethylene, as described in Annex 1. The displacements of the distal part of the humerus were fixed in all directions at a distance of 150 mm from the tip of the tuberosity while a load of 200 N was applied uniformly in the tendon free surface. The load direction is tangential to the terminal part of the supraspinatus model.

By pretensioning the spring acting on the suture models at a 40 N load level, we introduce the knot-tying effect that leads to an interface pressure >0 between tendon and bone and between sutures and tendon.

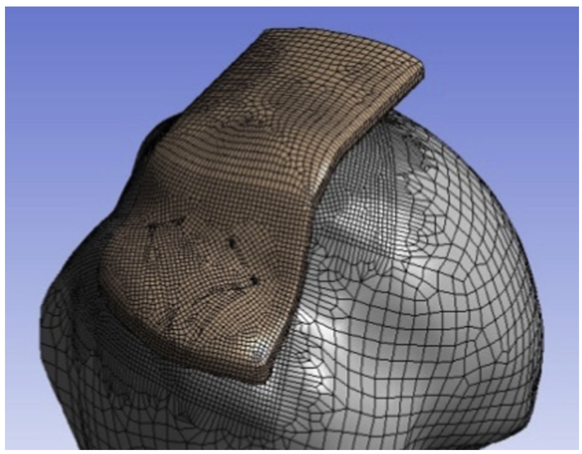

Figure 3 The 3D mesh used in the simulation.

The contact between the supraspinatus and bone has been treated as frictional, with a friction coefficient equal to 0.1; setting this option in ANSYS permits both parts to freely separate and slide and the contact to be modified by the suture effect, starting from the initial model at time 0 (with a preload level of 0 N). The load sequence is reported in the diagram of Figure 4.

To assess the contact area extension and the pressure distribution, we made use of an APDL (ANSYS Parametric Design

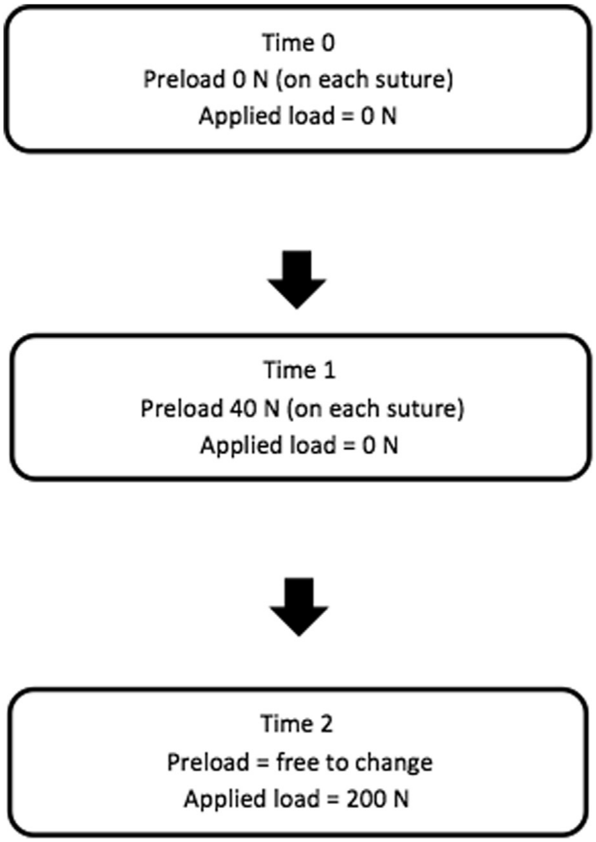

Figure 4 Load history adopted in the model.

Language) macro script; a threshold pressure level of 0.0001 MPa was adopted when the contact area was computed. The maximum accepted interpenetration between bodies in the contact area is 0.001 mm. The surface considered by the macro and computed as the contact area consists only of the real tendon-bone interface, excluding the fixing device volume. At time 0 between tendon and bone, a geometrical gap of 0.05 mm was created. This gap was eliminated when the spring was pre-tensioned, simulating knot-tying action. Once the external load is applied, the initial contact area can change because of the lack of a counteracting downward pressure able to guarantee the stability of the initial contact.

The pretension forces, acting on the springs connected to the various sutures, were evaluated on the basis of the pressure measured by Tuoheti at al.[30] The different models have been evaluated from a geometrical standpoint, and we measured what we named the repair area, which is the area with a positive contact pressure greater than the threshold value.

Results

Table II reports the real active repair area, which excludes the presence of synthetic material (the typical anchor diameter widely used in in vivo and ex vivo study spans between 5 and 6 mm) over which the attachment is not possible.

Table II Evaluated contact areas of the various techniques

Repair method	Repair area (mm^2)	Contact area with a positive pressure (mm^2)
Transosseous equivalent (4 anchors, 2 screwed and 2 impacted laterally)	103	42
Single row	35	15.9
Double row	87	26.8

The TE approach had a wider positive contact area. The SR technique produced by far the lowest footprint coverage compared with the other techniques. The DR technique provides an increase of the positive contact area equal to 69%, whereas the TE technique gives 164% more in comparison to the SR technique.

The area reported in the column "repair area" probably underestimates the real value, but the value reported in the last column represents the element area sum that displays a positive value of the contact pressure. By the macro described before, it is possible also to filter these data, excluding very low values (that cannot be considered of real effectiveness in a dynamic environment to prevent the tendon from sliding) and spurious peaks, which are more related to the model used instead of having a real physical meaning.

Figure 5 shows the representation of the contact area in the various constructs. What appears evident is the effect of the suture bridge configuration on the final computation of the contact area. It is evident how a different suture layout can significantly vary the final extension of the contact and the residual tension in the sutures, having a direct effect on the sliding resistance (Fig. 6).

Discussion

In this study, we presented a finite element method as an alternative to laboratory tests to compare various repair configurations. Even if the absolute values require a more extensive experimental validation (see Annex 1 for encouraging preliminary experimental results), we could consider this comparative approach a flexible tool that can be used to define the repair strategy supported by the biologic and mechanical factors that increase the probability of having an intact construct.

Previously reported findings[12,14,27,28,33] support finite element analysis as a promising tool to evaluate and to compare various repair configurations in an easy, fast, and flexible way. Indeed, mechanical factors are at the basis of a biologic healing process (mechanical stability, pressure distribution, contact area, reduction of local stress peaks).

In this study, biologic factors were not considered. The problem was addressed from a mechanical standpoint, and

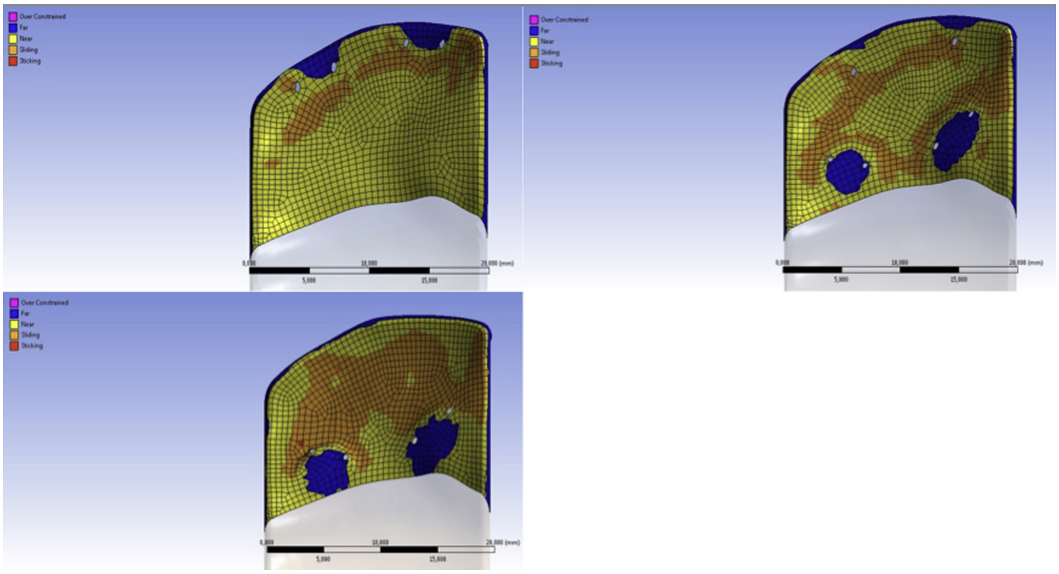

Figure 5 Qualitative maps of the supraspinatus-bone contact layout; *orange* represents the area in contact with a positive applied pressure. The free surfaces are in *yellow*, and the absence of any contact (device insertion areas) is shown in *blue*. *Upper left*, single row (SR); *upper right*, double row (DR); *lower left*, transosseous equivalent (TE).

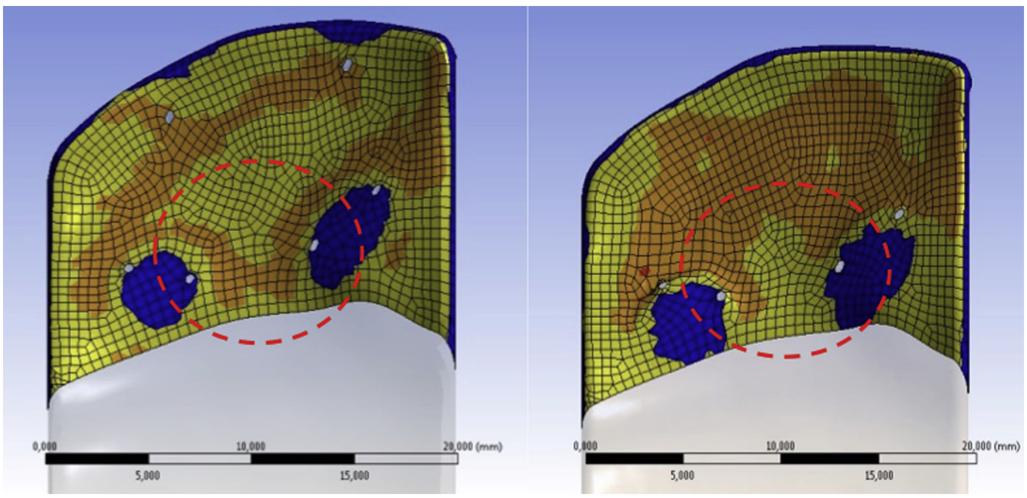

Figure 6 Direct comparison of how a "dead area" (*left side*, evidenced by the *red circle*) is transformed in a compressive area through the use of a bridging configuration (*right side*, evidenced by the *red circle*).

the configurations that maximize some geometrical and mechanical factors considered the basis of the healing process were discovered. These factors could be identified as area of contact, pressure and pressure distribution over the contact, avoidance of local peak stress, and reduction of the tension over the repair. Milano et al[19] have interestingly demonstrated how excess tension applied over the repair can significantly impair the biomechanical results. The biologic relation between applied pressure at the soft tissue–bone interface and the integrity of the repair has been demonstrated.[11]

Several studies evaluated repair integrity, which we consider of importance,[11,19,29] such as Duquin et al,[11] who compared mechanical stability and repair integrity.

Many papers have compared SR and DR techniques in terms of clinical outcomes, also looking at biomechanical and anatomic constructs.[5,7,10,11,31] Data collected in our study suggest how the SR approach gives several high-stress peaks in the areas close to the anchor's position; these pressures decrease sharply in the interanchor space.

Our study follows previously reported trends,[2,6,18-21,23,24] and highlights important aspects of repair techniques. Suture bridging between the various anchors or tunnels (in the case of a transosseous approach) appears to be essential to increase contact area. Suture bridges are effective not only for their load sharing effect that reduces local peaks, but also to make a "dead area" (defined as the footprint area between anchors that in some repair configurations presents a zero contact pressure) active in the repair zone.

Another interesting finding is the effect of humeral head shape on contact asymmetry, as it is evident that footprint shape strongly influences contact pressure. Reshaping by decortication will significantly increase compression in transosseous techniques, and maintain tight contact between soft tissue and bone. A stable construct can enlarge the contact pressure and normalize pressure distribution, which may help keep the repair intact. Our proposed method may maximize footprint area coverage to enhance repair integrity.

Following the guidelines provided by Viceconti et al,[32] we are aware of the approximations introduced in this model. However, the purpose of our study was to compare the efficacy of different repair techniques, and we do not feel the simplifications biased our conclusions.

There are limitations to this study which would require validating the results in an experimental setup; however, comparing similar results in the literature, ours follow the data trends, but the results are extremely dispersed.[2,6,18,20,25]

Our approach allows assessment of multiple variables, and seems promising. Dar[8] reported that statistical methods should also be implemented for a more comprehensive comparison of various techniques.

Conclusion

Our study confirms that DR and TE repairs lead to an increase of the contact area and to a better distribution of the pressure coverage. Although the finite element method is a theoretic one, the approach we presented could be a valid tool to predict and reproduce different configurations and to infer conclusions concerning different repair approaches. Further biomechanical studies are required to compare the repair techniques in this study.

Disclaimer

Pietro Garofalo is employed as Product Manager at NCS Lab Srl (Carpi, Modena, Italy).

Matteo Mantovani is Technical Director at NCS Lab Srl (Carpi, Modena, Italy).

The other authors, their immediate families, and any research foundations with which they are affiliated have not received any financial payments or other benefits from any commercial entity related to the subject of this article.

Supplementary data

Supplementary data related to this article can be found at http://dx.doi.org/10.1016/j.jse.2015.09.002.

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