MASS-SPRING MODELING OF THE EARDRUM FOR SURGICAL SIMULATION

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INTRODUCTION

The finite-element (FE) method was first utilized by Funnell and Laszlo to model the eardrum [1]. Since then it has been applied by numerous researchers in order to understand the normal function of the eardrum and middle ear and to investigate the effects of pathology and reconstructive surgery [2].

We are developing an interactive computer-based simulator using virtual-reality interfaces to train surgical residents in myringotomy [3], a procedure in which an incision is made in the eardrum as part of the treatment for ear infections. In the current implementation, all tissues are treated as static, and a line is simply drawn to visually indicate the incision. However, based on an end-user survey, it was concluded that deformation and splitting of the eardrum in response to a myringotomy blade provides useful visual information to trainees and needs to be incorporated into the simulator. Realistic deformations and real-time performance are required for interactive training applications like ours. Although FE models of the eardrum can be highly accurate, no existing models of the eardrum achieve real-time performance.

The mass-spring (MS) method has been used successfully in a number of interactive surgical simulators [e.g., 4]. In this method, the tissue or continuum to be modeled is represented by a distributed set of points connected by springs to simulate the elasticity of the structure. Real-time computations are possible; however, the accuracy of the method is uncertain. The objective of this work was to develop a MS model of the human eardrum and evaluate its accuracy in comparison to an FE model.

METHODS

Finite Element Model

The FE model used in this study is shown in Figure 1 and is based on an existing model [5] with the exception that the triangular thin-shell elements used previously were replaced by 4-noded quadrilateral thin-shell elements. The original model was based on shape data acquired from a cadaveric human ear using phase-shift shadow moiré topography, a non-contacting optical technique for the measurement of surface shape. The transfinite interpolation (TFI) technique was used to convert the triangular mesh to a quadrilateral mesh [6]. This technique maps a rectangular grid onto an arbitrary surface shape. Using the existing triangular mesh, the pars tensa, pars flaccida and manubrium were outlined. With these outlines, a two-dimensional mesh of the eardrum consisting of quadrilateral elements was generated using the TFI method. Depth information (i.e., z-coordinates) was inferred from the original triangular mesh by interpolation. The final quadrilateral mesh has 906 nodes and 816 elements. This is the minimum mesh resolution that results in convergence of calculated displacements.

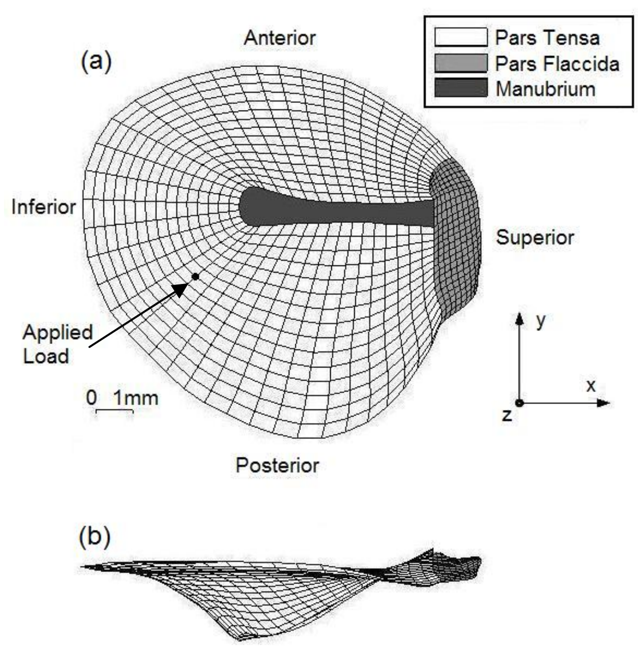

Figure 1: FE model of a human eardrum. (a) Lateral view as approximately seen through ear canal. (b) Postero-anterior view.

The pars tensa makes up the main surface of the eardrum and was modeled as linearly elastic, isotropic and homogeneous, assumptions that have been used in previous studies [7]. Under these assumptions, the pars tensa could be fully characterized with a single Young's modulus of 20 MPa [8] and a Poisson's ratio of 0.3 [1]. Its thickness was taken to be 110 µm across its entire surface [9]. The pars flaccida is a much smaller part of the eardrum that is superior to the pars tensa and not directly relevant to myringotomy simulation since all incisions are always made in the pars tensa. For modeling simplicity, its properties were assumed to be identical to that of the pars tensa. The manubrium is the portion of the malleus bone attached to the pars tensa. It was modeled as being completely immobile to simplify modeling by isolating eardrum response from the middle-ear ossicular chain and cochlear load; therefore, no elements were needed to model the manubrium [10]. The periphery of the eardrum was modeled as fully clamped [1]. The attachment of the eardrum to the manubrium was modeled as simply supported.

Mass-Spring Model

The MS model was derived from the FE model by treating the edge of each quadrilateral element as a spring. To represent shearing effects, diagonal springs were added as shown in Figure 2. This is a commonly used spring topology for simulating deformable structures such as cloth [11]. The stiffness of each spring was computed from the Young's modulus, Poisson's ratio and surface geometry using the formula proposed by Lloyd *et al* [12].

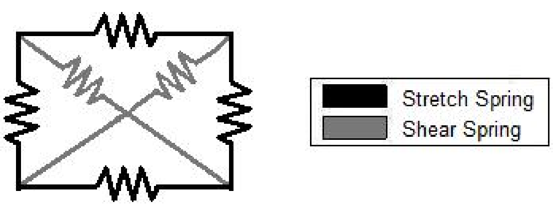

Figure 2: Spring configuration for one quadrilateral surface element.

Simulation

Simulations were performed using both the FE and MS models when applying a point force to the same single node; the point chosen is shown in Figure 1. The direction of the force was along the negative z-axis, simulating contact by a myringotomy blade. The magnitude of the force was varied from 1 mN to 100 mN. Although forces have not been measured during actual myringotomies, this range is likely higher than forces experienced during actual surgeries and provides a rigorous comparison of the two modeling approaches. The FE model was simulated using the Abaqus software (Simulia Inc., RI, USA), taking into account geometric nonlinearity. Such nonlinearity considers the changes in geometry as the structure is deformed by the application of force. The MS model was simulated using publicly available software [13].

RESULTS

Figure 3 shows iso-amplitude displacement contours computed using the FE model for a load of 50 mN. The displacement component shown is in the negative z-direction. As expected the maximum displacement occurs at the point of application of the force and decreases away from it.

Figure 4 shows iso-amplitude contours computed using the MS model. Qualitatively, the displacement pattern computed using the MS model is similar to that computed using the FE model. However, with the FE model the displacements decrease slightly more quickly with increasing distance from the point of application of the force than they do with the MS model. This is especially true for small displacements.

For other force magnitudes, the displacement patterns computed using the MS model agree well with those computed using the FE model with the exception of a small difference in displacement magnitude. To illustrate this difference, Figure 5 shows force-displacement curves computed using both models. The maximum displacement magnitude is shown. Below 50 mN, the maximum displacement computed using the MS model is larger than that computed using the FE model. The reverse is true above 50 mN. However both curves indicate nonlinear behavior with displacement magnitude increasing less than in proportion to the applied force as the force increases.

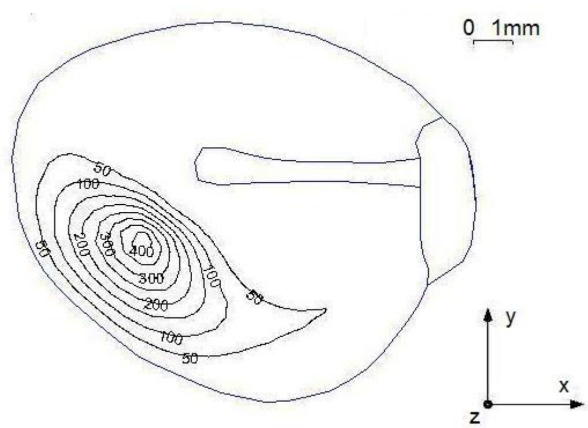

Figure 3: Iso-amplitude displacement contours computed using the FE model in response to a 50 mN point load. Only the magnitude of the z-direction component is shown. Displacements are in um.

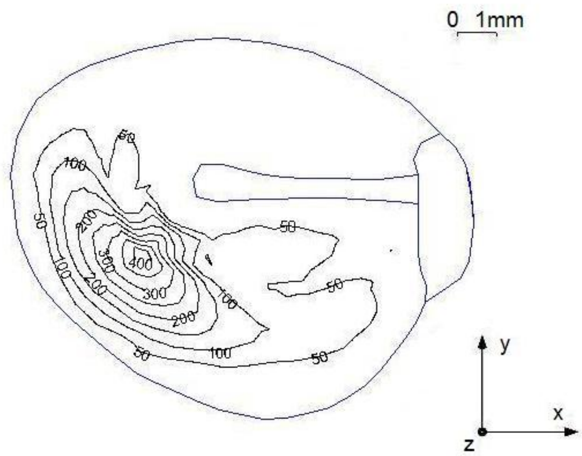

Figure 4: Iso-amplitude displacement contours computed using the MS model in response to a 50 mN point load. Only the magnitude of the z-direction component is shown. Displacements are in um.

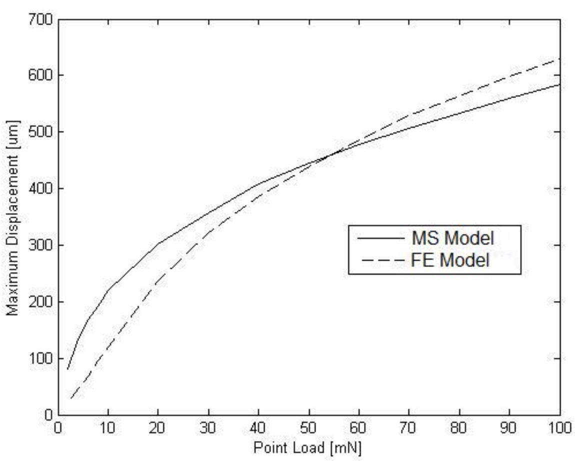

Figure 5: Maximum displacement magnitude as a function of the applied load magnitude.

Five minutes of computation time was required for each force application when using the FE model. By comparison, 30 seconds were required to compute a solution using the MS model.

DISCUSSION

MS models produce displacement patterns that are qualitatively similar to those computed using FE models, although there are slight differences in overall displacement magnitudes. This mismatch might be caused by the fact that the MS model as implemented ignores bending stiffness and only accounts for membrane stiffness, whereas the FE model accounts for both. For small applied forces (below 50mN in the current study), it is likely that stresses induced by bending dominate membrane stresses, resulting in smaller displacements computed using the FE model relative to the MS model. However, it is not clear if the differences in computed displacements would be visually distinguishable to users of interactive training simulators such as ours.

Computations done using the MS model used in this work are clearly faster than those done using the FE model. No special code optimization was used when implementing the MS computation software. Real-time performance (i.e., computation rates of 30 Hz or faster) are easily achieved with code optimization. It is worthwhile to note that real-time performance is now achievable with FE models as well, for example, by using graphics processing units [14]. The MS method has the added advantage of simplicity of formulation which leads to ease of implementation.

Several simplifications have been used in constructing both models. For instance, the Young's modulus and thickness are assumed to be constant across the surface of the eardrum. Measurements at a few locations on real eardrums suggest that there is some variation in both the Young's modulus [15] and thickness [16]. Moreover, the fibrous ultrastructure of the eardrum suggests that it is better modeled as being orthotropic. Furthermore, although geometric nonlinearity is a minimal requirement for modeling large deformations of the eardrum (i.e., displacements comparable to its thickness) [10], material nonlinearity may also need to be considered if strains become large. Since the goal of this work was simply to compare the MS method to the well established FE method, detailed modeling of inhomogeneity, anisotropy and nonlinearity was not required. However, it may be necessary to refine the MS model further if such refinements result in differences in displacement patterns that are visually discernable.

The eardrum models were also further simplified by considering the manubrium to be immobile which obviates the need to model the loading of the eardrum by the middle-ear ossicular chain and cochlea. It also simulates a common pathology (fixation of the ossicular chain). However, modeling the middle ear would be necessary when attempting to simulate a mobile ossicular chain. For the present study, the focus was on comparing FE and MS models of the eardrum, and modeling the rest of the middle ear was not necessary for this comparison.

CONCLUSION

Our results show clear differences between finite element and mass-spring models of the eardrum. This is not surprising, as past work has shown an exact match between these two modeling techniques may be difficult. However, for interactive training purposes, an exact visual match may not be necessary in order to adequately teach the skills required for myringotomy. In the range of deformation seen by surgeons during this procedure, the differences between the two modeling techniques are visually small.

ACKNOWLEDGEMENTS

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