

HUMAN BODY FE MODEL REPOSITIONING: A STEP TOWARDS POSTURE SPECIFIC – HUMAN BODY MODELS (PS-HBM)

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ABSTRACT

This study reports a methodology to generate anatomically correct postures of existing human body finite element models while maintaining their mesh quality. The developed method is based on computer graphics techniques and permits control over the bone kinematics followed. Initial mesh quality of the model being repositioned is quantified and maintained during the process in order to preserve the suitability of the model for subsequent use. For some critical areas (soft tissues around knee joint like capsule, menisci) where element distortion is large, mesh smoothing technique is used to improve the mesh quality. The tool developed repositions a given lower extremity model (knee joint) in just a few seconds.

Keywords: Human body FE model, Posture, Repositioning, Mesh morphing

IN THE LAST DECADE, many human body finite element models (THUMS (Maeno, T. et al. 2001), HUMOS2 (Vezin, et al. 2005) and JAMA/JARI (Sugimoto, T. et al. 2005) etc.) have been being developed. However, the geometry of most of these FE models is limited only to standard occupant or pedestrian postures. Whereas, in real life, the body can be in various postures such as, standing, walking, running or jogging postures and out-of-position (OOP) occupant postures. Compromises due to non-availability of FE models for different postures may lead to erroneous conclusions and may limit the use of these models. On the other hand developing FE models for all possible limb positions is not viable. Therefore, personalization of existing FE models to get Posture Specific Human Body Models (PS-HBM) through limb adjustments needs to be done.

Very few studies report repositioning techniques for human body FE models. Vezin et al., (2005), report that, the HUMOS2 has been equipped with posture change capability. Two methods have been described for repositioning the model. In the first approach a database of pre-calculated positions is being used and intermediate positions are obtained by linear interpolations between nearby positions. The second approach is based on interactive real-time calculations. However, they do not provide enough information about the technique and the quality of the results obtained; it is thus not possible to judge the accuracy of the anatomical relation among the body segments, the time required for repositioning and the quality of the mesh obtained.

Parihar (2004), repositioned the lower extremity of the THUMS model from an occupant posture to a standing (pedestrian) posture using a series of dynamic FE simulations. The upper leg was restrained and force was applied to the tibia. In each step the lower leg was given a rotation of 5 – 6 degrees. The results of the iterations were dependent on the constraints and contact interfaces defined as well as the accuracy of the geometry and the material properties. They reported that the simulation time was very long (about 72 Hrs for 90 degrees of flexion on an Intel P IV 2.4 GHz processor with 2 GB RAM) and required a large number of iterations and modifications. One more disadvantage of the method is that there is no direct control over the kinematics being followed. The positional accuracy of the repositioned model solely depends on the geometry, the contacts defined and the boundary conditions imposed.

Jani et al., (2009), have also reported repositioning of the human body FE model using FE simulations. They have investigated the time required for repositioning, control over the kinematics followed, and anatomical correctness of the repositioned (with respect to the final bone position) model and the level of user intervention needed. They have also concluded that the process requires

large CPU time and does not permit enough control over the kinematics followed. Also, while anatomical correctness of the final posture is questionable mesh quality of the repositioned model is poor and needs subjective mesh editing.

In the present study a method to reposition human body FE model has been proposed. The method is independent of model geometry and needs minimal subjective interventions. Also, available data (kinematics of bones, strains in soft tissues etc.) can easily be incorporated. The method is based on standard computer graphics techniques like morphing and affine transformations. These techniques are widely used for generating human body animations. Many researchers have used various morphing approaches to generate animations of human-like characters (Sheepers et al., 1997, Aubel 2001, Dong et al., 2002, Blemker et al., 2005 and Sun et al., 2000). Most of these approaches have been developed to handle surface models or models created using ellipsoids (to represent muscles) and can therefore not be used directly. So, a new method based on Delaunay Triangulation was developed to handle soft tissues while repositioning.

METHODS

MODEL GEOMETRY: The focus of the current study is on the knee joint repositioning. Hence, lower extremity model (excluding pelvis and foot) was used in the present study. The geometry model was extracted from a full human body FE model (the General Motors (GM) / University of Virginia (UVA) 50th percentile male FE model (Untaroiu et al., 2005 and Kerrigan et al., 2008). [a]

[b]

Fig. 1 (a) and (b) below show the external and detailed view of the model in its initial configuration. The lower extremity of this model consists of 45 components. The bones (femur, tibia, fibula and patella) are modelled in multiple layers with a shell mesh for cortical and hexahedral mesh for spongy bone. The model also includes ligaments (modelled with solid elements) of the knee joint, menisci (modelled with solid elements), tendons (modelled with shell elements), knee capsule (modelled with shell elements), and muscles (modelled with solid elements).

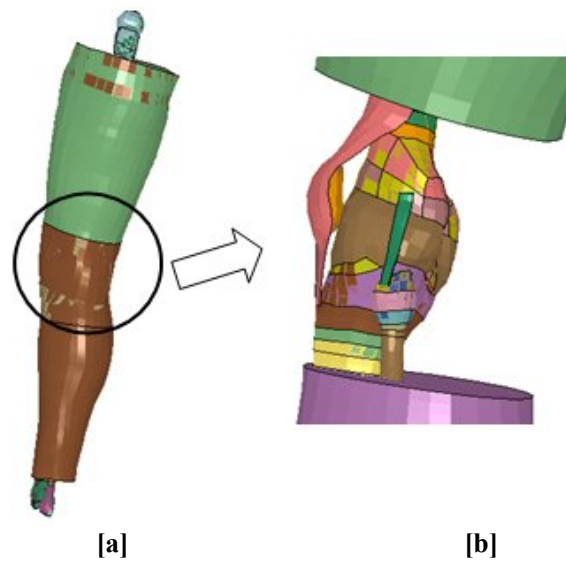


Fig. 1 Lower Extremity Model

Important landmarks like femur head, greater trochanter, lesser trochanter, linea aspera, condyles, epicondyles on the femur and condyles, intercondylar eminence, Gerdy's tubercle, tibial tuberosity and malleolus on the tibia were located. These landmarks were used for defining reference points during the mapping operation.

In order to achieve accurate positioning of the model, it is essential that accurate information of the kinematics of bones constituting the joint be used. In the present study the focus is on the knee joint which involves two distinct motions: Tibiofemoral motion and Patellofemoral motion. These motions are discussed here.

TIBIOFEMORAL MOTION: There has been extensive work on the kinematics of tibiofemoral joint. Various techniques like, CT scans with biplanar image matching (Asano et al., 2001, 2005) and MRI scans (Hill et al., 2000, Martelli et al., 2002, Freeman et al., 2005, Pinskerova et al., 2001, Johal et al., 2005) have been used to study the kinematics. Besides these techniques use of fluoroscopy, X-rays radiographs and Radio-Stereometric Analysis (RSA) has also been reported.

These techniques have been used both in living knee weight bearing (Li, 2007, Johl, 2005, Hill, 2000, Asano, 2001, Asano, 2005) and cadaveric conditions (Elias, 1990, Hollister, 1993, Churchill, 1998, Iwaki, 2000, Most, 2004, McPherson, 2005). These studies have reported description of the tibial and femoral articular surfaces and their relative movement in various coordinate systems and references.

Even though a qualitative agreement is observed among data reported with regard to type of motion, a quantitative comparison on knee joint kinematics might be difficult between different studies due to the different activities studied as well as different coordinate systems used to measure knee kinematics.

It is understood from literature that the tibiofemoral motion is the combination of two movements: (1) rotation of femur with respect to tibia (or vice versa) about a flexion-extension axis (**F-E axis**) (referred here as pure flexion-extension rotation) and (2) sliding / translation of femoral condyles over the tibial plateau.

Conventionally, it is believed that tibiofemoral flexion-extension rotation has “no fixed” axis i.e. flexion-extension rotations occurs around an instantaneous axis which sweeps a ruled surface (known as an axode, Fig. 2). This eventually makes it difficult to describe or reproduce anatomically correct pure flexion-extension rotation in knee kinematic studies using mathematical models. However, studies have shown that F-E axis can be better described while considering it as fixed (Hollister, 1993, Churchill, 1998, Stiehl, 1995). Recently, Eckhoff et al. (2003) have also demonstrated that the knee F-E axis can be approximated by a single cylindrical axis in posterior femoral condyles. Considering these findings from literature, a fixed (single) F-E axis has been located and used in the present study.

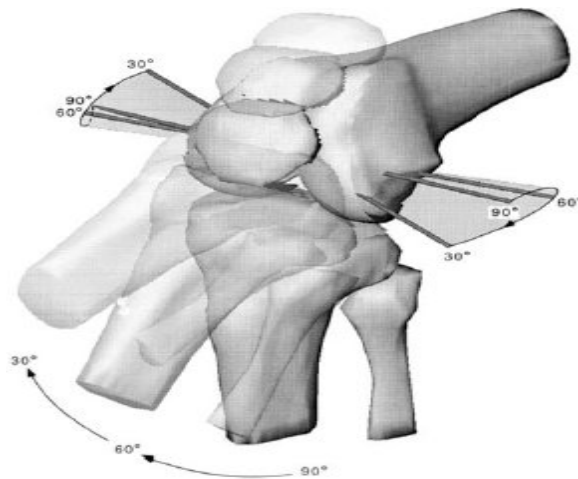


Fig. 2 The axode describing the knee joint motion (Adapted from Mow et al., 2000)

There have been different views about the orientation (definition) of F-E axis in the posterior femoral condyles. Two definitions of fixed F-E axes are widely used.

1. Transepicondylar axis (TEA) defined as axis connecting most prominent points on the lateral and medial femoral condyles or axis connecting femoral origins of lateral ligaments (Blankevoort et al., 1990, Hollister et al., 1993, Churchill et al., 1998, Miller et al., 2001). The axis has also been used in total knee arthroplasty (TKA) (Berger et al., 1993, 1998).
2. Geometric centre axis (GCA) defined as the axis passing through medial and lateral centers of the circular profiles of the posterior condyles (Asano et al. 2001, 2005, Eckhoff et al., 2001, Pinskerova et al., 2001). The GCA has also been used to represent the posterior geometry of

the femoral condyle (Eckhoff et al., 2001) and kinematic data (Blankevoort et al., 1990; Freeman, 2003; Hill, 2000; Iwaki et al., 2000). The GCA is shown in Fig. 3 (a).

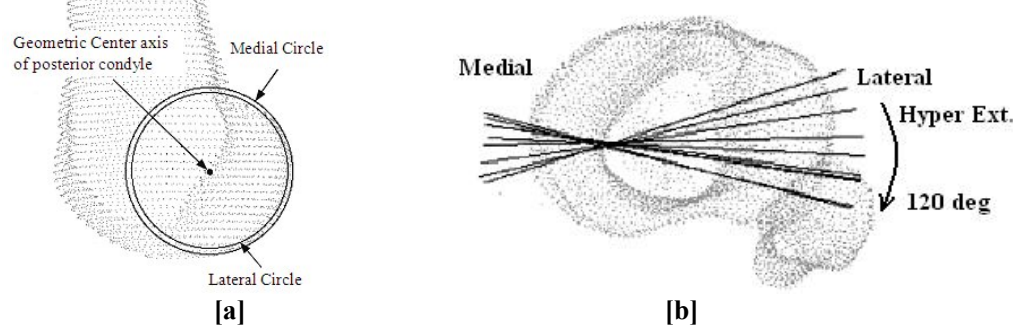


Fig. 3 (a) Circles fitted on posterior femoral condyles and GCA (b) Movement of GCA with flexion indicates external rotation of femur (Adopted from Asano et al., 2001)

Most et al. (2004), have analysed the sensitivity of the knee joint kinematics calculation to selection of flexion axes (TEA or GCA) and established that as long as a clear definition of the flexion axis is given any of the axes can be used to describe knee joint kinematics. The GCA has been used as the F-E axis in this study.

Researcher of knee kinematics have established that the medial femoral sliding (femoral translation on the tibial plateau) is minimal (typically in the range of ± 1.5 mm Iwaki et al., 2000) while the lateral femoral sliding is large (could be as large as 24 mm, Iwaki et al., 2000). This fact can also be observed in a typical movement of GCA from hyper extension to 120 degree flexion (Fig. 3 (b)). This fact is also observed by many others [Churchill et al., (1998), Asano et al. (2001, 2005), Pinskerova et al., (2001), Wretenberg et al., (2002), Johl et al., (2005)] and it has been suggested that this uneven femoral sliding can be approximated as an external rotation of femur about longitudinal rotation axis (LR axis) fixed in the medial compartment of the tibia.

From this discussion about knee kinematics, the followings can be concluded:

1. Tibiofemoral motion can be approximated by two rotations about two axes. (a) The Flexion about F-E axis followed by longitudinal rotation about LR axis.
2. Flexion extension axes can be approximated by a fixed stationary axis in the posterior femoral condyles.

Implementation of Tibiofemoral kinematics: In the present study tibiofemoral motion data from Asano et al., (2001) has been used. The F-E axis was located in the posterior femoral condyles, passing through centres of the circles approximating the lateral and medial posterior femoral condyles. The radii of these circles were 20.52 mm on the lateral side and 23.31 mm on the medial side, which are within the range of 18 – 23 mm and 20 – 25 mm respectively as reported by Pinskerova et al. (2001). Also, the lateral and medial ends of this axis were found to be within the area of femoral origins of lateral and medial collateral ligaments respectively.

The longitudinal axis was located between a point on the medial tibial plateau (approximate centre of the contact path) and centre of the tibio-talar joint. These two axes have been shown in the Fig. 4. Rotation of the femur about these two axes approximates the knee joint motion.

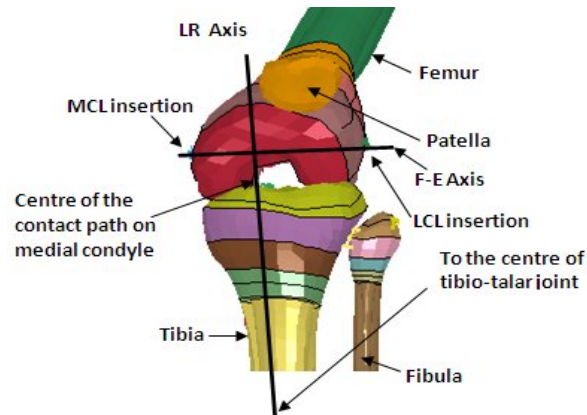


Fig. 4 The Flexion Extension axis and Longitudinal Rotation axis

Mechanical and anatomical axes of the femur and tibia were located as shown in Fig. 5. Tibiofemoral flexion was measured as the angle between the long axes (anatomical axes) of the tibia and the femur, projected on the sagittal plane as suggested by Li et al., (2007).

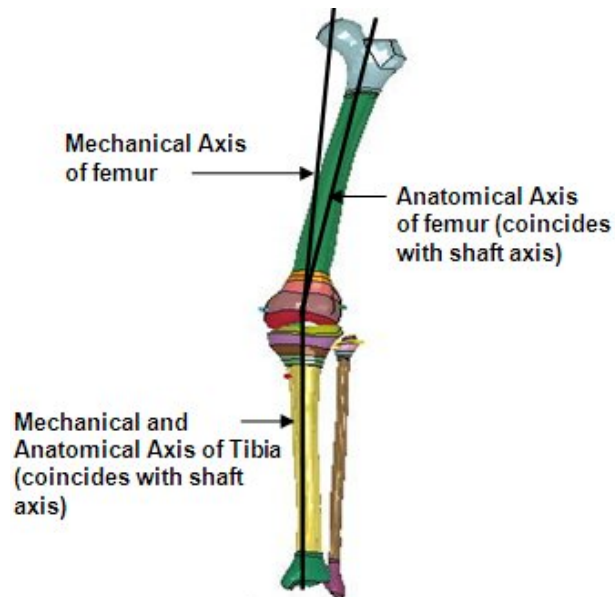


Fig. 5 Mechanical and Anatomical axes of tibia and femur

In the algorithm developed, femur, tibia and fibula components were treated as rigid bodies. For generating flexion, femur was subjected to rotation about F-E axis followed by rotation about LR axis while keeping the tibia fixed. The relation between these two rotations (Table 1) was adopted from Asano et al., (2001). Flexion angles were verified as angles between projections of anatomical axes of tibia and femur projected on sagittal plane. Thus, complete control over the bone kinematics followed was achieved.

Table 1 Relation between F-E and LR Axes rotation (Asano et al., 2001)

Flexion Angle (Degrees)	External Rotation of femur about LR axis Mean $\pm$ sd (Degrees)
Hyperextension	- 4.2 $\pm$ 1.7
0	0
15	7.6 $\pm$ 2.6
30	13.3 $\pm$ 2.2
45	16.6 $\pm$ 2.7
60	19.5 $\pm$ 4.1
75	21.2 $\pm$ 4.6
90	22.9 $\pm$ 4.9
105	24.9 $\pm$ 4.1
120	23.8 $\pm$ 4.8

PATELLOFEMORAL MOTION: Like tibiofemoral joint, there are many studies of the patellofemoral joint that investigate various aspects of its motion. Some studies have focused on analysis of patellofemoral forces (contact areas / pressures) (Singerman et al., 1995, Zavatsky et al., 2004), while few others consider patellofemoral kinematics (Heegard et al., 1995, Asano et al., 2003, Zavatsky et al. 2004, Li et al., 2007). Even the data of kinematics is presented either in terms of translations (Anterior-Posterior, Medial-Lateral, Superior - Inferior) of the patellar origin with respect to the tibial reference (Li et al., 2007) or femoral reference (Asano, et al., 2003) and rotations (Li et al., 2007, Zavatsky et al., 2004).

Accurate patellar tracking, and definition of normal tracking, have not yet been achieved in either experimental or in clinical conditions (Katchburian et al., 2003). Furthermore, no universal agreement exists on the definition of normal patellar tracking (Grelsamer et al., 2001).

In the present study data from Zavatsky et al. (2004) has been used. In his study, patellar flexion, internal – external rotation and medial – lateral tilt are measured against the tibiofemoral flexion. The coordinate system used for the patella is shown in Fig. 6. Patella was subjected to affine transformations in accordance with this data.

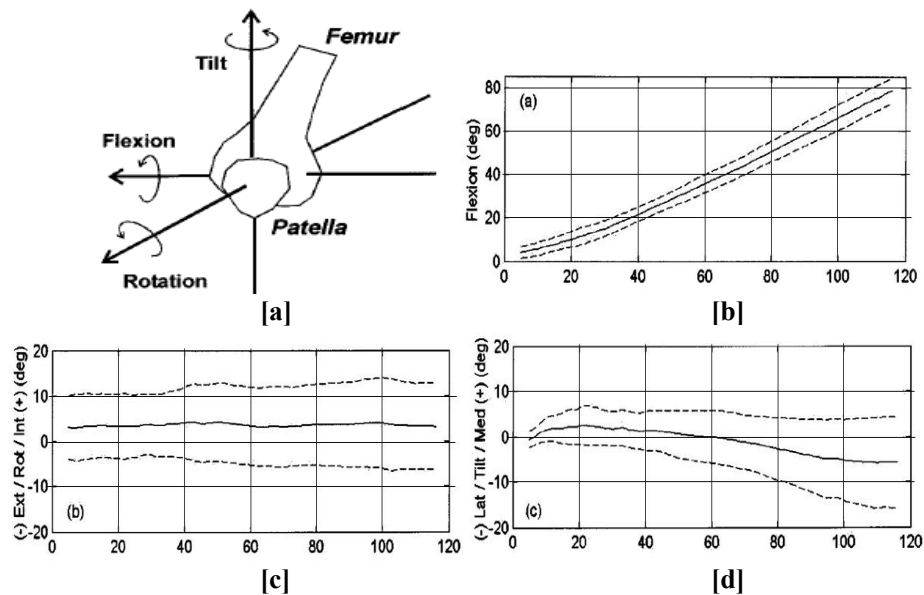


Fig. 6 (a) Coordinate system for patellar motion (b) Patellar flexion (c) External – Internal rotation of patella (d) Lateral - Medial tilt, (Adopted from Zavatsky et al., 2004)

MORPHING: Morphing or metamorphosis is a widely used computer graphics technique of deforming graphical objects (computer graphics models). The technique is used either to deform a given object intuitively through a Graphical User Interface (GUI) or to transform a given source object into a target object. Various techniques have been developed by graphic designers to deform human like structures for animation / games (Sheepers et al., 1997, Aubel 2001, Dong et al., 2002, Blemker et al., 2005 and Sun et al., 2000). But most of them either dealt with surface models defined with parametric surfaces or with structures defined by ellipsoids. In the present work a new technique has been developed to transform the model under consideration which is a composite model of shell and solid elements. Thus morphing was used in the present study to handle deformation of soft tissues after the bones were repositioned using affine transformations.

Implementation of Morphing: A code using VC++ (programming language and GUI) and OpenGL (Graphics platform) has been developed to handle both the processes viz., affine transformations of bones and morphing of soft tissue. This can be understood from the flow chart in Fig. 7.

First skin contours were identified in the initial configuration of the given model. Planes (total 86) were then determined for each contour of skin (section) as shown in Fig. 8. The space between the skin and outer surface of the bones was discretized into tetrahedrons (≈ 40000 tetrahedrons) using Delaunay triangulation. The soft tissue nodes were mapped into these tetrahedrons. This information was later used to map the soft tissues into the final position. The bones were given affine transformations as per the predetermined data. The skin contours were then transformed.

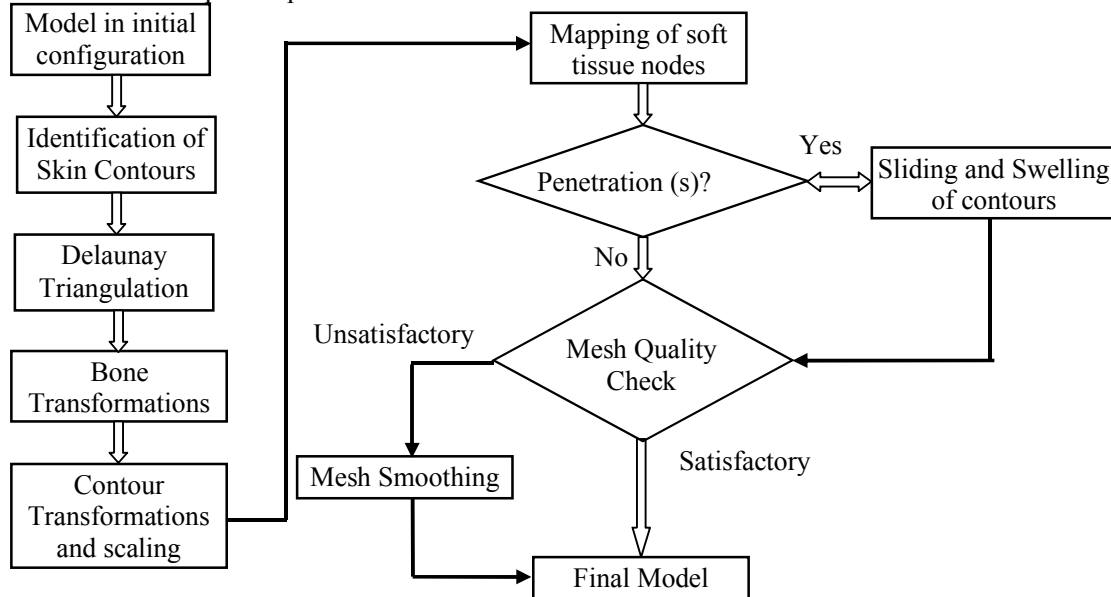


Fig. 7 Morphing process chart

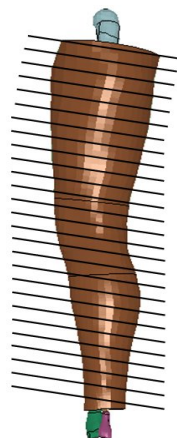


Fig. 8 Planes of skin contours

To transform the contours of the skin, if these contours are given an affine transformation with bones, they will penetrate into one another as shown in Fig. 9 (a)

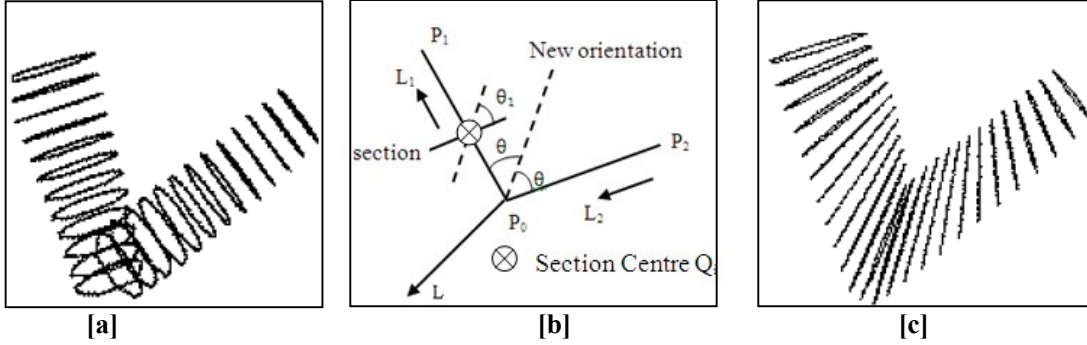


Fig. 9 (a) Penetrating contours (b) Plane rotation to remove penetration of contours (c) Corrected position of contours, (Jianhua et al., 1994)

This penetration was removed by rotating each layer about an axis lying in the plane of the layer in the medial-lateral direction. The amount of rotation was calculated as shown in Fig. 9 (b). The link P_1P_0 corresponds to the femoral axis and link P_2P_0 to the tibial axis. P_0 is the knee joint position. L_1 and L_2 are inferior – superior directions of upper leg and lower leg respectively. Suppose the angle between two links (joint angle) is 2θ . When the joint angle changes, the attached section (one such section is shown) first undergoes joint local transformations defined by the joint hierarchy as indicated by solid line in Fig. 9 (b). The orientation of the plane at the joint was set to be the bisection plane of the joint. The normal of i^{th} planes were then set as follows:

For each plane from P_1 to P_0 ,
 {

Obtain centre of rotation Q_i .

Find the axis of rotation L of plane as $L_1 \times L_2$.

Adjust plane normal N_i by rotating about axis L with rotation angle given by:

$$\theta_{rot,i} = \theta * \sin\left(\frac{\pi * (i-1)}{2 * N_{upper}}\right) \quad \text{for } i=1, 2, \dots, N_{upper} \quad (1)$$

$$\theta_{rot,i} = -\theta * \sin\left(\frac{\pi * (i-1)}{2 * N_{lower}}\right) \quad \text{for } i=1, 2, \dots, N_{lower} \quad (2)$$

where,

N_{upper} = number of planes on upper link,

N_{lower} = number of planes on lower link

}

To prevent volume squashing, contours were scaled out so that distances of points on these contours to the attached link remain constant. Suppose after transformations a point A (in Fig. 10) is located on the i^{th} contour with center at Q_i . After rotations of cross-section about axis of rotation passing through Q_i (L in Fig. 9 (b)), the point A is transformed to A_1 . In initial posture, the distance of point A to the link is r_0 . We scale point A_1 out to position A_2 in the direction A_1Q_i for approximately constant volume.

$$scale_factor = \frac{r_0}{dist(A_1, P_0P_1)}, \text{ Hence} \quad (3)$$

$$A_2 = Q_i + scale_factor \times (A_1 - Q_i) \quad (4)$$

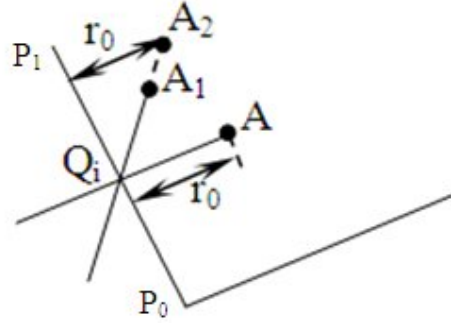


Fig. 10 Prevention of volume squashing

Once the final position of bones and skin nodes is achieved, the nodes of the tetrahedrons generated in stage “A”, are already relocated. Nodes of the soft tissues are inside the tetrahedrons and were then mapped to the new position using a linear mapping in their respective tetrahedrons. Thus, the repositioned model is obtained. As soft tissues in the model are mapped with Delaunay triangulation which is a complete partition of the soft tissue space, it is expected that there are no penetrations in the repositioned model (unless initial model has penetrations). But due to complex geometry of the model and the fact that some bone elements come in contact with some soft tissues only after transformations (for instance, elements of condyles are not exposed to the capsule and flesh initially, came in contact after flexion) penetrations were observed. These penetrations were removed by a combination of linear sliding and swelling of contours calculated on the basis of the maximum penetration.

The repositioned model was finally checked for mesh quality parameters like maximum aspect ratio, maximum warpage, maximum skew and minimum jacobian. The model is then subjected to mesh smoothing, if needed.

RESULTS AND DISCUSSION

The model repositioned with the method developed is shown in Fig. 11. The method has permitted complete control over the kinematics of the bones and also maintained the mesh quality to a predefined level. At this stage, the flexion up to 90 degrees has been implemented.

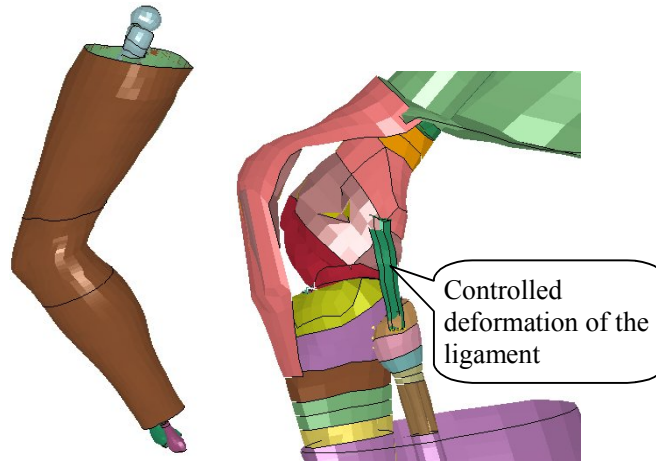


Fig. 11 (a) Repositioned model (b) Detailed view of knee joint

Unlike, repositioning based on FE simulations (Parihar et al., 2004, Jani et al., 2009), where the repositioning takes a considerable amount of time, the method developed in the present study an existing model can be repositioned in few seconds. Also, the method presented does not require any precalculated positional data as needed in the tool developed by Vezin et al., (2005).

The comparison of various mesh quality parameters has been given in the Table 2 below.

Table 2 Comparison of mesh quality parameters

Quality Parameter	Initial Configuration	Repositioned Model	Repositioned Model (with mesh smoothing)
Aspect Ratio			
Warpage			
Skew			
Jacobian			

The method does not require any remeshing after repositioning and the model in the new posture can readily be used for subsequent simulations. Further, there is no change in the mesh during repositioning and the element connectivity and element / node numbering remains unchanged.

CONCLUSION

A methodology to update an existing human body FE model available in a given posture (occupant or pedestrian) to posture specific FE model has been presented in the study. Though the methodology presented has been applied to the knee joint only, it is believed that it can easily be extended to other joints like hip joint and shoulder joint. The prime objective of the study was to develop a methodology to reposition limbs to achieve an anatomically correct position, at the same time maintaining the mesh quality of the initial model.

References:

1. Asano, T., Masao, A., Tanaka, K., Tamura, J. And Nakamura, T., (2001), In vivo three dimensional knee kinematics using a biplanar image matching technique. Clin Orthop, 388, 157-166.
2. Asano, T., Akagi, M. Nakamura, T., (2005), The functional Flexion-Extension axis of the knee corresponds to the surgical epicondylar axis, J Arthroplasty, 20 No 8, 1060-1067.
3. Aubel, A. and Thalmann, D., (2001), Interactive modelling of the human musculature, In Proc. of Computer Animation.
4. Blankevoort, L., Huiskes, R., de Lange, A., (1990), Helical axes of passive knee joint motions. Journal of Biomechanics, Vol. 23, Issue 12, pp. 1219–1229.
5. Blemker, S. S. and Delp, S. L., (2005), Three dimensional representation of complex muscle architectures and geometries, Annals of Biomedical Engineering, 33, No. 5, 661-673.
6. Churchill, D.L., Incavo, S.J., Johnson, C.C., Beynon, B.D., (1998), The transepicondylar axis approximates the optimal flexion axis of the knee. Clin Orthop 356, 111–118.
7. Dong, F. and Clapworthy, G. J., (2002), An anatomy based approach to human muscle modelling and deformation, IEEE Visu and Comp Gr, 8, 2, 154-170.
8. Elias, S. G., Freeman, M.A.R., Gokcay, E. I., (1990), A correlative study of the geometry and anatomy of the distal femur, Clin Orthop, 260, 98 -103
9. Eckhoff, D. G., Bach, J. M., Spitzer, V. M., Reinig, K. D., Bagur, M. M., Baldini, T. H., Rubinstein, D. And Humphries, S., (2003), Three dimensional morphology and kinematics of the distal part of the femur viewed in virtual reality- Part II, J. Of Bone and Joint Surg., Vol. 85-A, Suppl. 4, pp. 96-104.
10. Eckhoff, D. G., Dwyer, T. F., Bach, J. M., Spitzer, V. M. and Reinig, K. D, (2001), Three dimensional morphology and kinematics of the distal part of the femur viewed in virtual reality- Part I, J Bone Joint Surg Am. 2001; Vol. 83, Suppl 2., pp. 43-50.
11. Freeman, M. A., Pinskerova, V., (2003), The movement of the knee studied by magnetic resonance imaging. Clinical Orthopaedics and Related Research, No. 410, pp. 35–43.
12. Freeman, M. A. R. and Pinskerova, V., (2005), The movement of the normal tibio-femoral joint, J. Biomech, 38, 197-208.
13. Grelsamer, R. P. and Weinstein, C. H., (2001), Applied biomechanics of the patella, Clin Orthop, No. 389, pp. 9-14.
14. Heegaard, J., Leyvraz, P. F., Curnier, A., Rakotomanana, L., Huiskes, R., 1995, The biomechanics of human patella during passive knee flexion, J. Of Biomechanics, Vol. 28, No. 11, pp. 1265-1279.

15. Hill, P. F., Vedi, V., Williams, A., Iwaki, H., Pinskerova, V. and Freeman, M. A. R., (2000), Tibiofemoral movement 2: the loaded and unloaded living knee studied by MRI, *J B Joint Su*, 1196 -1198.
16. Hollister, A.M., Jatana, S., Singh, A.K., Sullivan, W.W., Lupichak, H., 1993. The axes of rotation of the knee. *Clin Orthop* 290, 259–268.
17. Iwaki, H., Pinskerova, V., Freeman, M.A.R., (2000), Tibiofemoral movement 1: the shapes and relative movements of the femur and tibia in the unloaded cadaver knee: studied by dissection and MRI, *J Bone Joint Surg* 82B, 1189– 1195.
18. Jani, D., Chawla, A, Mukherjee, S., Goyal, R. and Nataraju, V., (2009), Repositioning the human body lower extremity FE models, *SAE # 2009-01-0922*.
19. Katchburian, M. V., Bull, A. M., Shih, Y. F., Heatley, F. W., and Amis, A. A., (2003), Measurement of Patellar Tracking: Assessment and Analysis of the Literature, *Clin. Orthop. Relat. Res.*, 412, pp. 241–259.
20. Johal, P., Williams, A., Wragg, P., Hunt, D. and Gedroyc, W., (2005), Tibio-femoral movement in the living knee, A study of weight bearing and non-weight bearing knee kinematics using ‘interventional’ MRI, *J. Biomech*, 38, 269-276.
21. Kerrigan, J., Parent, D., Untaroiu, C., Crandall, J., and Deng, B., (2008), A new detailed multi-body model of the pedestrian lower extremity: Developed and preliminary validation, *IRCOBI 2008*.
22. Li, G., Papannagari, R., Nha, K. W., Defrate, L. E., Gill, T. J. and Rubash, H. E., (2007), The coupled motion of femur and patella during In vivo weightbearing flexion, *J. of Biomech. Engg.*, ASME, Vol. 129, pp. 937-943.
23. Maeno, T., and Hasegawa, J., (2001), Development of a finite element model of the total human model for safety (THUMS) and application to car-pedestrian impacts, *The 17th ESV Conference*.
24. Martelli, S. Pinskerova, V., (2002), The shapes of the tibial and femoral articular surfaces in relation to the tibiofemoral movement, *J B Joint Su*, 48, 4, 607-613.
25. McPherson, A., Karrholm, J., Pinskerova, V., Sosna, A., Martelli, S., (2005), Imaging knee motion using MRI, RSA/CT and 3D digitization, *J. Biomech*, 38, 263-268.
26. Miller, M. C., Berger, R. A., Petrella, A. J., Karmas, A., Rubash, H. E., (2001), Optimizing femoral component rotation in total knee arthroplasty. *Clin. Orthop.*, No. 392. 38-45.
27. Most, E., Axe, J., Rubash, H., and Li, G., 2004, Sensitivity of the Knee Joint Kinematics Calculation to Selection of Flexion Axes, *J. Biomech.*, Vol. 37, Issue 11, pp. 1743–1748.
28. Mow, V.C., Flatow, E.L., Ateshian, G.A. (2000) Biomechanics. In: *Orthopaedic Basic Science*, J. A. Buckwalter, T.A. Einhorn, and S.R. Simon (eds), American Academy of Orthopaedic Surgeons, Rosemont, IL, 2nd ed, pp. 133-180.
29. Parihar, A., (2004), Validation of human body finite element models (knee joint) under impact conditions, *M. Tech. Thesis*, Indian Institute of Technology Delhi.
30. Pinskerova, V., Iwaki, A., and Freeman, MAR., (2001), The shapes and movements of the Femur and tibia in the unloaded cadavric knee: A study using MRI as an anatomical tool, Chapter 10. In: Insall, J., Scott, W.N. (eds.), “*Surgery of the Knee*”, 3rd Ed. Saunders, Philadelphia, USA, pp. 255–284.
31. Ramsey, D.K., Wretenberg, P.F., (1999), Biomechanics of the knee: methodological considerations in the in-vivo kinematic analysis of the tibiofemoral and patellofemoral joints. *Clinical Biomechanics*, Vol. 14, 595–611.
32. Sheepers, F., Parent, R., Carlson, W. And May, S. F. (1997), Anatomy based modelling of the human musculature. *SIGGRAPH '97*, 163-172.
33. Singerman, R., Berilla, J., and Davy, D. T., (1995), Direct in vitro determination of the patellofemoral contact force for normal knees, *J. Biomech. Eng. ASME*, Vol. 117, pp. 8–14.
34. Stiehl, J. B., Abbott, B.D., (1995), Morphology of the transepicondylar axis and its application in primary and revision Total Knee Arthroplasty, *J Arthroplasty*, 10, 785-789.
35. Sugimoto, T. and Yamazaki, K., (2005), First results from the JAMA human body model project, *The 19th International ESV Conference*.
36. Sun, W., Hilton, A., Smith, R. And Illingworth, J., (1999), Layered animation of captured data, Animations and Simulations'99, 10th Eurographics workshop proceedings, 145-154.
37. Untaroiu C., Darvish K., Crandall J., Deng B., Tai Wang J., (2005), A Finite Element Model of the Lower Limb for Simulating Pedestrian Impacts, *Stapp Car Crash Journal*; Vol. 49.

38. Vezin, P. and Verriest, J. P., (2005), Development of a set of numerical human models for safety, The 19th International ESV Conference.
39. Wretenberg, P., Ramsey, D.K., Nemeth, G., (2002), Tibiofemoral contact points relative to flexion angle measured with MRI. *Clinical Biomechanics*, Vol. 17, Issue 6, pp 477–485.
40. Zavatsky, A. B., Oppold, P. T., Price, A. J., (2004), Simultaneous In vitro measurement of patellofemoral kinematics and forces, *J. Biomech. Eng. ASME*, Vol. 126, pp. 351-356.
41. Jianhua, S., Magnenat Thalmann, N., Thalmann, D., (1994), Human Skin Deformation from Cross Sections, *Proc. Computer Graphics International '94*, Melbourne, Australia.