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Repositioning Human Body Lower Extremity FE Model

Dhaval Jani, Anoop Chawla, Sudipto Mukherjee, Rahul Goyal
Department of Mechanical Engineering, Indian Institute of Technology Delhi, India

V. Nataraju
General Motors India Pvt. Ltd

ABSTRACT

This study aims to develop a methodology to generate anatomically correct postures of existing human body finite element models while maintaining their mesh quality. This repositioning is often done by running dynamic simulations. Such simulations, while taking a lot of time have the disadvantage of giving distorted elements as well as require a lot of expertise and have subjective interventions. Also, the anatomical correctness of the final position, and the kinematics followed during repositioning by dynamic simulations are uncertain. The developed method is based on computer graphics techniques and repositions a joint in just a few seconds. Repositioning of the lower extremity was also carried out using Finite Element (FE) simulations and analysed. The repositioning results from the two techniques were compared and it was found that the technique based on computer graphics gave satisfactory results.

INTRODUCTION

Of late, 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.) are 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. Compromises due to non-availability of FE models for different postures may lead to erroneous simulation results 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. Parihar (2004), repositioned the lower extremity of the THUMS model from a occupant posture to a standing (pedestrian) posture. The leg position was modified using a series of FE simulations. The upper leg was restrained and force was applied to tibia. In each step 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 defined. 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.

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

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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.

The present study describes a quick repositioning technique and compares the method with FE simulations for repositioning. The time required for repositioning, control over the kinematics followed, anatomical correctness of the repositioned (with respect to the bone position) model and the level of user intervention needed are compared. A lower extremity FE model has been used and the knee joint is repositioned to study suitability of the methods in flexion extension motions.

A method based on standard computer graphics techniques like morphing and affine transformations has been developed. 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 a. 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). Therefore it is difficult to adopt any approach without modifications for repositioning human body FE models. So, after studying various techniques of morphing and deformation, a new method was developed to handle soft tissues while repositioning.

MATERIALS AND METHODS

To achieve the objective of repositioning using computer graphics techniques the model components were first segregated into two groups, viz., rigid components (bones) and deformable components (soft tissues). The rigid components were then given affine (rigid) transformations while the deformable components were morphed. Finally, the mesh quality and penetrations were checked and improved wherever needed. For the repositioning using simulations, angular velocity about the transepicondylar axis was applied to the tibia while keeping the femur fixed.

MODEL GEOMETRY

The focus of the current study is on the knee joint repositioning. Hence, a lower extremity model (excluding pelvis and foot) was used in the present study. The geometry was extracted from a full human body FE model (Figure 1) (the General Motors (GM) / University of Virginia (UVA) 50th percentile male FE model (Untaroiu et al., 2005 and Kerrigan et al., 2008).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).

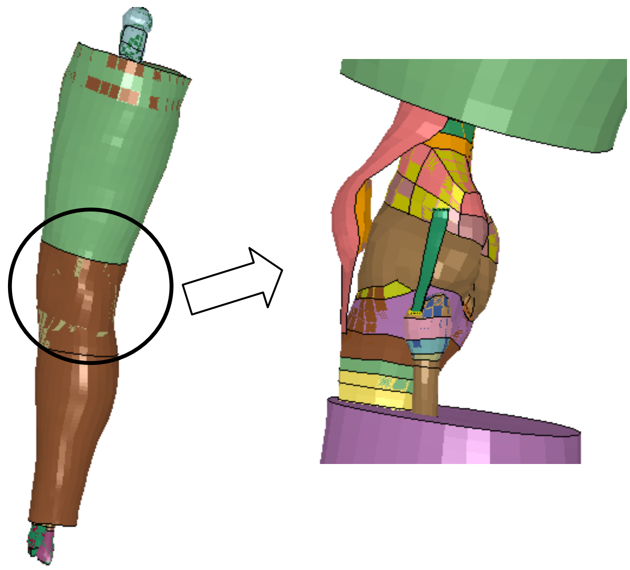


Figure 1 Lower Extremity model (Untaroiu et al., 2005)

Mechanical and anatomical axes of the leg were located in the femur as well as the tibia as shown in Figure 2.

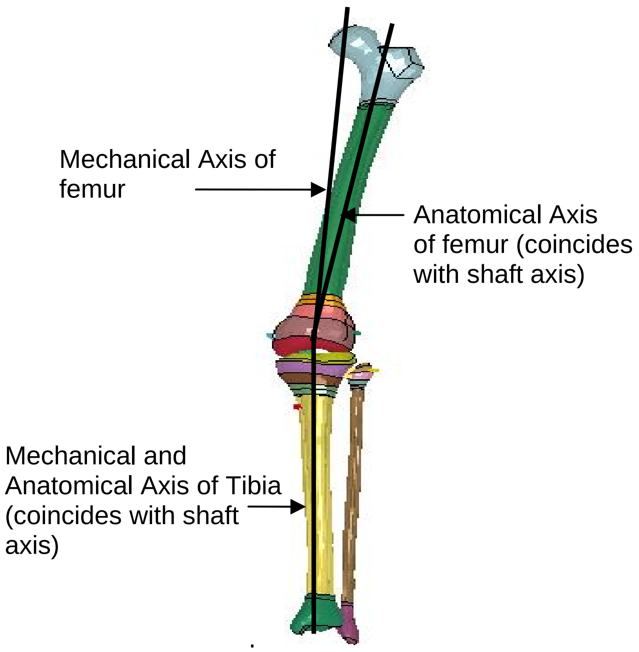


Figure 2 Mechanical and Anatomical Axes of Femur and Tibia

These were later used to define the flexion-extension angle. Also important landmarks like femur head, greater trochanter, lesser trochanter, linea aspera, condyles, epicondyles on the femur and condyles, intercondyler eminence, Gerdy's tubercle, tibial tuberosity and malleoulus on the tibia were located. These landmarks were used to define a mapping between the bone and the soft tissues.

TIBIOFEMORAL MOTION

In this section, tibiofemoral kinematics has been discussed. Location and orientation of the flexion extension axis (F-E axis) and longitudinal rotation axis (LR axis) are deduced. Figure 3 shows the orientation of these axes.

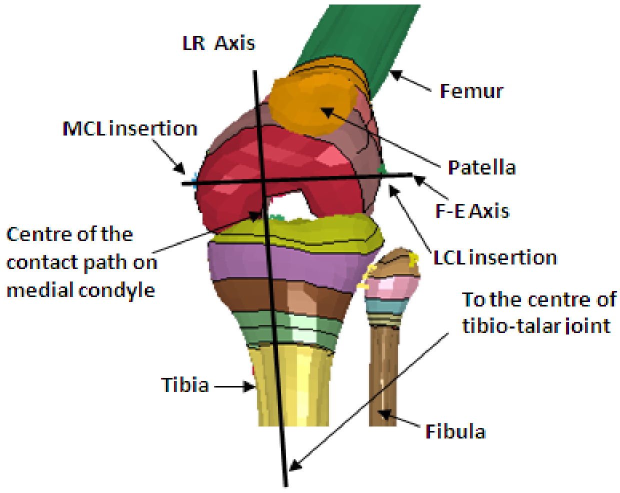


Figure 3 Orientation of Flexion Extension (F-E) and Longitudinal rotation (LR) axes

A lot of literature is available describing tibiofemoral kinematics. In order to achieve accurate positioning of the model, it is essential that accurate information of the tibiofemoral motion should be used.

Various techniques have been used to study tibio-femoral kinematics. These include the use of CT (Asano et al., 2001) and MRI scans (Hill et al., 2000, Martelli et al., 2002, Freeman et al., 2003, Pinskerova et al., 2004, Johal et al., 2005). Besides these techniques use of fluoroscopy, X-rays radiographs and Radio-Stereometric Analysis (RSA) has also been reported.

Researchers have defined different coordinate systems to describe tibiofemoral motion. Grood, et al. (1983) have concluded that, calculations of angular motion in knee kinematics can vary depending upon the coordinate system used. Variations in locating a coordinate system can cause motion around one axis (say flexion/extension) to be interpreted in part as being around another axis (say longitudinal axis). This error is known as kinematic cross-talk (Ramsey et al., 1999; Piazza et al., 2000; McPherson et al., 2005, Freeman et al., 2005).

Tibiofemoral motion is agreed to be a combination of rolling and sliding. The rolling takes place about the flexion-extension axis while sliding of femoral condyles occurs over the tibial condyle. The rotation about the flexion extension axis is a complex motion as the axis of rotation is not a single static axis i.e. it does not remain fixed, but it is an instantaneous axis. Indeed, the axis sweeps a ruled surface (an axode) as the knee flexes (Figure 4) (Mow et al., 2000).

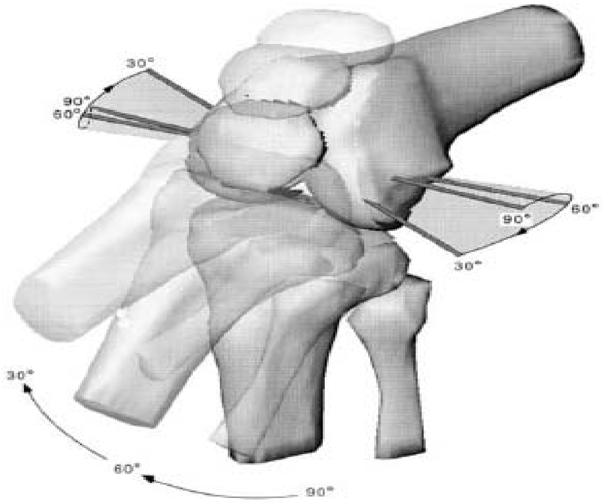


Figure 4 The axode describing the knee joint motion (Adapted from Mow et al., 2000)

The tibiofemoral motion can be approximated by two simultaneous rotations about two axes (Hollister et al., 1993, Churchill et al., 1998). The first rotation takes place about F-E axis and the second rotation is about LR axis. However, there has been no common agreement among researchers about the exact location of the longitudinal axis, even though there is agreement that this axis lies in the medial side of the tibia.

The rotation about the longitudinal axis gives effect of internal external rotation about medial condyler interface between tibia and femur. Also, this secondary rotation is responsible for apparent translation / sliding of the lateral femoral condyle over the lateral tibial condyle. Figure 5 (Adapted from Iwaki et al., 2000) shows, amount of translation of the femoral condyles over the tibial condyles during flexion and extension. It can be seen that translation or sliding of femoral condyles over tibial condyles is unequal i.e. the medial condyle translates by a small amount (approximately 3 mm) compared to the lateral condyle (approximately 23 mm).

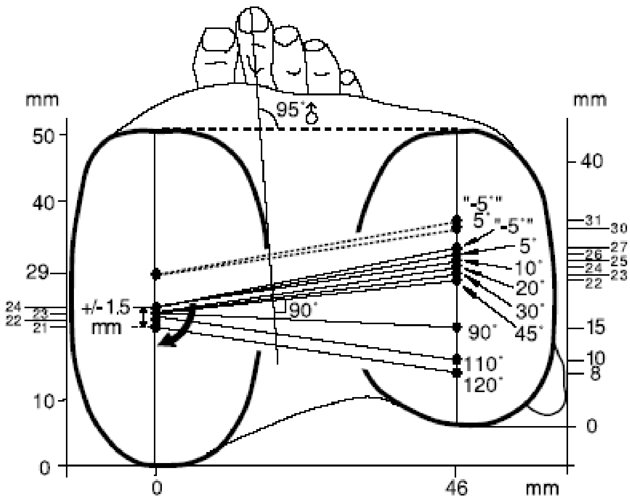


Figure 5 Translation of femoral condyles over tibial condyles at different flexion angle (Adapted from Iwaki et al., 2000)

Also, over the complete range of motion this sliding is not linear, making the motion more complex to reproduce. To reproduce this motion with simultaneous rotations, it is important to establish the relation between the amount of flexion and simultaneous rotation to be carried out about LR axis. The authors have adopted results of Asano et al., (2001) for the relation between these two rotations. These results are also complementing other studies (Pinskerova et al.,2001; Iwaki et al., 2000; Hill et al.,2000; McPherson et al., 2005).

The exact location and movement of the flexion extension axis varies from person to person. The posterior femoral condyle is known to be circular in shape [Pinskerova, et al., (2001), Asano et al. (2001), and Hollister et al. (1993)]. Asaon et al., (2001) has found that, the arc of posterior femoral condyle in the sagittal plane consisted of the distal femur articulated with the tibia from approximately 30^{o} to 120^{o} knee flexion. Therefore, the axis passing through the centres of the circles fitting the condyles approximately can be used as a fixed flexion extension axis. Elias et al. (1990), have also reported a fixed flexion extension axis in the posterior femoral condyles. Churchill et al. (1998), Stiehl et al. (1995) and Asano et al. (2005) have also identified a single fixed axis, the transepicondylar axis, which can be used to approximate flexion extension axis.

Therefore, the knee motion has been defined by two rotations, one about the F-E axis and second about the LR axis. Since the axis is being specified with respect to the bones, the choice of the coordinate system is not significant in the present study. Further, even though a static approximation of the F-E axis is being used, the method is general and can also be implemented if the axode representing the change in F-E axis is used.

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

1. Knee motion can be approximated by rotation about two axes. (a) The Flexion extension axis (b) the Longitudinal rotation axis.

2. Flexion extension axes can be approximated by a fixed stationary axis passing through centres of the posterior condyler circles (the transepicondylar axis - passes through insertion of lateral ligaments).

3. Longitudinal rotation axis lies in the medial compartment of the tibia.

MORPHING

Metamorphosis or morphing finds wide applications in computer graphics, animations and engineering. The technique is used to deform a given graphical object with both geometric information and graphical attributes (colour, texture, shading etc.). A deformation algorithm is developed and applied to a graphical object. The technique transforms the object in to another object. The process can be understood by an example given in the Figure 6 which shows a square being transformed to a circle.

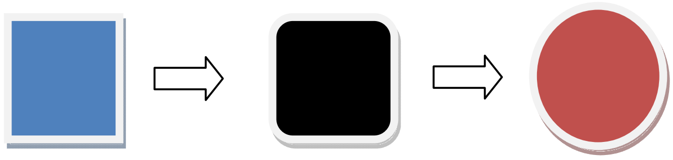


Figure 6 Morphing

The morphing algorithm first rounds the sharp corners and then adjusts the edges to create a circle.

In the present study the morphing technique has been used for handling the deformation of the soft tissues after the bones are transformed.

REPOSITIONING OF HUMAN BODY FE MODEL

Two methods for the repositioning of the human body FE model have been developed in this study. This section describes both the techniques.

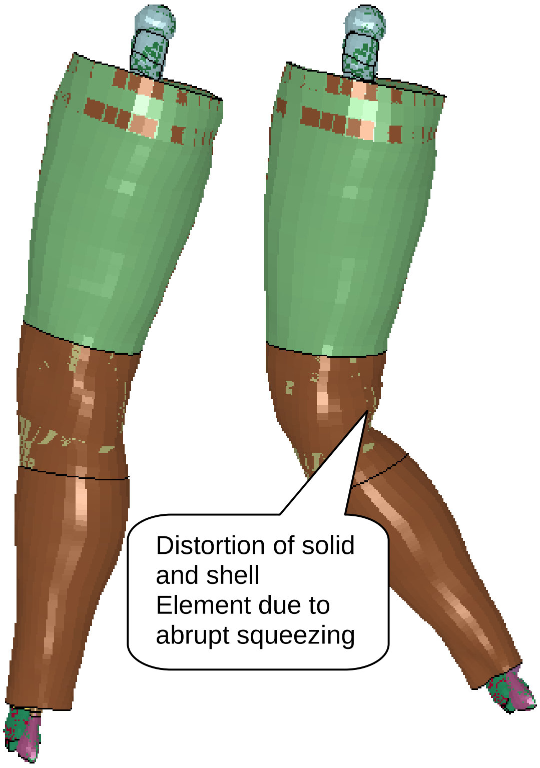
Repositioning using FE simulations

This section describes a simulation based approach to reposition the knee joint. The results are later compared with those from the morphing based approach developed in this study. The simulation setup, boundary conditions and findings of the study are discussed below.

Simulation Setup

It was found that in the available configuration, the model represents approximately 8o knee flexion condition as shown in Figure 7 (a). The available model had a maximum warpage of 196.94, maximum aspect ratio of 10.76 and a minimum Jacobian of 0.21.

For an anatomically correct repositioning of body parts it is essential to establish appropriate axes at joints about which rotation is to be carried out. The transepicondylar axis (through insertion of lateral ligaments) was located on the model and chosen as flexion extension axis. Also, circles to approximate Flexion Facet (Iwaki et al., 2000) were generated considering their centre on the F-E axis. The flexion radii of these circles were 20.52 mm on lateral side and 23.31 mm on the medial side, which are within the range of 18 – 23 mm and 20 – 25 mm respectively reported by Pinskerova et al. (2001). The longitudinal axis was located between points on the medial condyle of tibia and centre of tibio-talar joint. Rotation of the femur about these two axes will approximate the knee joint motion. In the simulations, rotation is defined only about the flexion extension axis and it is expected that rotation about LR axis will occur due to the anatomical structure of the model, and the forces generated due to the contacts defined at the knee joint.



(a) (b)

Figure 7 (a) Initial configuration (b) Flexed knee

Boundary Conditions

The tibia was fixed by constraining all its degrees of freedom. Femur was given an angular velocity of 0.5 rad/s about the flexion extension axis identified earlier.

Results of simulation

The CPU (Intel Core2Quad 2.4 GHz processor Q6600, 8 GB DDR2 RAM) time for the simulation was around 4 hrs to achieve about 90 degrees of flexion. The flexed leg (after 8 ms) is shown in Figure 7(b). As can be seen the elements of the skin and soft tissues in the knee region are distorted severely and hence mesh quality in these regions became quite poor (Maximum Warpage= 179.9, Maximum Aspect Ratio= 27.94, Minimum Jacobian = -119.98).

Deformation of Lateral Collateral Ligament (LCL), knee capsule, Anterior Cruciate Ligament (ACL) and Posterior Cruciate Ligament (PCL) is also similar. This is not desired. And the biofidelity of the repositioned model becomes doubtful.

Repositioning using Affine transformations and Morphing

The requirements of a positioning tool can be stated as follows:

1. *Anatomical correctness of the repositioned model:* The repositioned model should have an anatomically correct position. As discussed in the previous section, repositioning of lower extremity (Flexion - Extension) using FE simulations doesn't give the anatomically correct position.

2. *Good Mesh quality:* The repositioned model should have a good mesh quality. This is important for its use in subsequent simulations.

3. *Low processing time:* Time required for the repositioning process should be low.

In addition, as shown in Figure 8 the Medial Collateral Ligament (MCL) gets severely distorted.

Keeping these issues in mind a method for repositioning the knee joint FE model has been developed, which is based on techniques used in computer graphics and animation, namely, affine transformations and morphing.

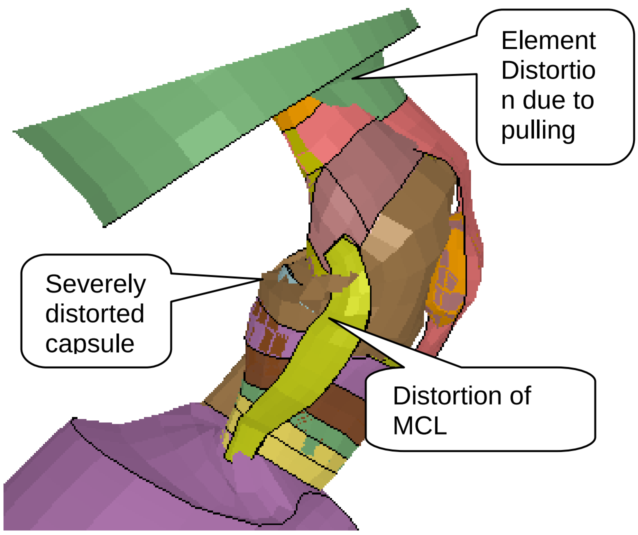


Figure 8 Detailed view of flexed knee

The model components have been categorized in two groups, viz, rigid components (bones) and deformable components (soft tissues).

Rigid components were transformed using an affine rotation transformation as they maintain geometric and graphical attributes of the object. On the other hand, soft tissues were transformed using morphing.

Thus, the femur was given a rotation about the flexion extension axis followed by a rotation about the longitudinal rotation axis to capture the external rotation of the femur. This way a known movement was achieved. The magnitude of the rotation about the LR axis has been taken as per data in Table 1. Intermediate rotations are linearly interpolated.

For transforming the soft tissues, a morphing based technique has been developed and implemented in VC++ and OpenGL.

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±SD (Degrees)
Hyperextensio	- 4.2 ± 1.7
0	0
15	7.6 ± 2.6
30	13.3 ± 2.2
45	16.6 ± 2.7
60	19.5 ± 4.1
75	21.2 ± 4.6
90	22.9 ± 4.9
105	24.9 ± 4.1
120	23.8 ± 4.8

The technique establishes a mapping of the space between the bones and the skin. This is used to regenerate the nodes of the soft tissues in the repositioned state. Using this technique, the repositioning of the lower limb to a 45^{o} flexion requires around 4 seconds. The output of the code is the repositioned model.

RESULTS AND DISCUSSION

In order to upgrade existing human body FE model to a PS-HBM two methodologies were evaluated. In both the methods the same F-E axis was used.

In the repositioning with FE simulations, it was expected that rotation about LR axis should occur automatically (as it happens in a human knee) during the simulation. But it didn't occur leaving the results anatomically incorrect. To full fill this condition of rotation about LR axis, one may need to either refine the mesh of condyles or perform multiple sequential simulations. This would further increase the time needed for the operation.

Initial results of repositioned model through computer graphics techniques are quite encouraging. The technique requires much lesser time compared to FE simulations. Moreover, it allows a much better control over the kinematics being followed by the bones and soft tissues. The problems of the distorted elements persist in this approach also but it is being taken care of by implementing mesh smoothing techniques.

Results obtained from this technique are shown in Figure 9.

As can be seen in the Figure 9, deformation of soft tissues is more reasonable than that obtained from simulations. Hence it is expected that such PS-HBMs will have a better biofidelity. The results can be further improved by using mesh refinement techniques which are still being implemented.

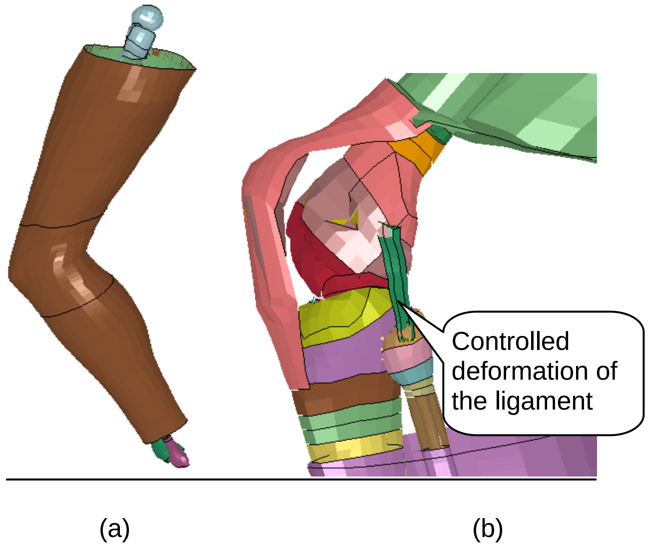


Figure 9 (a) Model repositioned with morphing and affine transformations (b) Detailed view of knee joint

CONCLUSIONS

In this study, results of repositioning of human body FE model using FE simulation were compared with those obtained from morphing. Repositioning using FE simulation is time consuming and produces anatomically incorrect model with poor mesh quality. Indeed, as the method doesn't require any kinematic input / constrain, repositioned models produced using the method does not reflect a correct kinematics also. Thus the method does not appear to be suitable for repositioning. On the other hand, results obtained from the morphing based techniques seem to be encouraging and are consistent with anatomical requirements as well as mesh quality requirements. Techniques for mesh smoothing / refinement are being implemented to improve the quality of distorted elements.

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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. 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.
5. 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.
6. 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.
7. 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
8. Freeman, M. A. R. and Pinskerova, V., (2005), The movement of the normal tibio-femoral joint, J. Biomech, 38, 197-208.
9. Grood, E.S., Suntay, W.J., (1983), A joint co-ordinate system for the clinical description of three-dimensional motions: application to the knee. J. Biomech Eng. 105, 136–144.
10. 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.
11. 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.
12. 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.
13. 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.
14. 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.
15. 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.
16. 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.
17. 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.
18. 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.
19. Parihar, A., (2004), Validation of human body finite element models (knee joint) under impact conditions, M. Tech. Thesis, Indian Institute of Technology Delhi.
20. Piazza,S.J.,Cavanagh, P.R., 2000. Measurement of the screw-home motion of the knee is sensitive to errors in axis alignment. J. Biomech 33, 1029–1034.
21. Pinskerova, V., Iwaki, A., and Freeman, MAR., (2001), The shapes and movments 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.
22. 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. 14, 595–611.
23. Sheepers, F., Parent, R., Carlson, W. And May, S. F. (1997), Anatomy based modelling of the human musculature. SIGGRAPH '97, 163-172.
24. Stiehl, J. B., Abbott, B.D., (1995), Morphology of the transepicondylar axis and its application in primary and revision Total Knee Arthroplasty, J Arthoplasy, 10, 785-789.
25. Sugimoto, T. and Yamazaki, K., (2005), First results from the JAMA human body model project, The 19th International ESV Conference.
26. Sun, W., Hilton, A., Smith, R. Amd Illingworth, J., (1999), Layered animation of captured data, Animations and Simulations'99, 10th Eurographics workshop proceedings, 145-154.
27. Untaroiu C., Darvish K., Crandall J., Deng B., Tai Wang J., A Finite Element Model of the Lower Limb for Simulating Pedestrian Impacts, Stapp Car Crash Journal; Nov 2005; 49.
28. Vezin, P. and Verriest, J. P., (2005), Development of a set of numerical human models for safety, The 19th International ESV Conference.