



REPOSITIONING HUMAN BODY FE MODELS – AND A CASE STUDY FOR THE LOWER EXTREMITY MODEL

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**REPOSITIONING HUMAN BODY FE MODELS – AND A CASE STUDY
FOR THE LOWER EXTREMITY MODEL**

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Objective: Human body finite element models (FE-HBMs) are available in standard occupant or pedestrian postures. There is a need to have FE-HBM in the same posture as a crash victim or to be configured in varying postures. Developing FE models for all possible positions is not practically viable. The current work aims at getting Posture Specific Human Lower Extremity Model by reconfiguring an existing one.

Methodology: A graphics based technique has been developed to reposition the lower extremity of a FE-HBM under flexion-extension. Elements of the model were segregated into rigid (bones) and deformable components (soft tissues). The bones were rotated about the flexion-extension axis and then about the longitudinal rotation axis to capture the twisting of the tibia. The desired knee joint movement was thus achieved. Geometric heuristics were then used to reposition the skin. A mapping was defined for the space between bones and the skin to regenerate the soft tissues. Mesh smoothing was then done to maintain mesh quality.

Results: The developed method permits control over the kinematics of the joint and maintains the initial mesh quality of the model. For some critical areas (in the joint vicinity) where element distortion is large, mesh smoothing is done to improve mesh quality.

Conclusions: A method to reposition human body FE model has been developed. The applicability of the method is demonstrated on the lower extremity (knee joint) of a human body FE model. The tool repositions a given model from 9° flexion to 90° flexion in just a few seconds and does not require subjective interventions. As the mesh quality of the repositioned model is maintained to a predefined level (typically to the level of model in initial configuration) the model is suitable for subsequent simulations and re-meshing is not needed.

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

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INTRODUCTION:

In the last decade, many FE-HBMs (THUMS (Maeno and Hasegawa 2001), HUMOS2 (Vezin and Verriest 2005) and JAMA/JARI models (Sugimoto and Yamazaki 2005) etc.) have been developed. The geometry of most of these models is limited only to standard occupant or pedestrian postures. However, in real life the body can be in various postures such as standing, walking, running or jogging postures and out-of-position occupant postures. The amount of muscle tensing, location of organs with respect to the vehicle and residual stress state of the musculoskeletal components depends on the posture acquired. 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 through limb adjustments needs to be done.

Very few studies reported repositioning techniques for FE-HBMs. Vezin and Verriest (2005) and Bidal et al. (2006), reported incorporation of posture change capability in HUMOS2. Two methods have been described by Vezin and Verriest (2005) for the repositioning. In the first approach a database of pre-calculated FE model positions are 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 load 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° 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 body 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 FE-HBM using dynamic finite element simulations. They have investigated the time required for repositioning, control over the bone kinematics, anatomical correctness of the repositioned model (with respect to the final bone position) and the level of user intervention needed. They have concluded that the process requires large CPU time and does not permit control over the bone kinematics without subjective interventions. Also, while anatomical correctness of the final posture is questionable, mesh quality of the repositioned model is poor and needs subjective mesh editing. They concluded that there is a need to develop another method to address the limitations of FE simulation based repositioning. Based on preliminary work it was also suggested that, in the interest of efficiency and efficacy, graphics based techniques would perform better.

In the present study, a new method to reposition FE-HBM has been proposed. The method is based on graphics techniques like morphing and affine transformations which are widely used for animating graphical characters. The technique also incorporates available data on kinematics of bones.

METHODS

This section describes the methodology developed for the FE-HBM repositioning along with the implementation for the lower extremity (knee joint).

Methodology:

The methodology is shown in the flow chart shown in Fig. 1. The process starts with segregation of model components into two groups: (1) rigid components (bones) and (2) deformable components (soft tissues). For a model segment (for instance, limb) to be repositioned, joint configuration (rotation or translation) and axes of motion at the joint are determined from literature (if available). The bones are repositioned with affine transformations, the skin is repositioned using graphics based heuristics and the soft tissues are mapped in the space between the bones using a Delaunay triangulation (Preparata and Shamos. 1988) based mapping to move them to a new position. Penetrations are controlled by swelling of the skin while mesh quality is improved (if necessary) through mesh smoothing. Though the methodology is generic and can be implemented to reposition any joint in a FE-HBM, this paper addresses its implementation on the lower extremity (knee joint).

Model Geometry:

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As the knee joint has been taken to demonstrate the repositioning methodology presented, a lower extremity FE model excluding pelvis and foot was used. The geometry of the 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). The model in the initial configuration is shown in Fig. 2 (a) and detailed view of the knee joint of model is shown in Fig. 2 (b). The bones (femur, tibia, fibula and patella) are modelled in multiple layers with shell elements representing the cortical part and hexahedral elements representing the spongy (trabecular) part. The model also includes ligaments (modelled with hexahedral elements) of the knee joint, menisci (modelled with hexahedral elements), patellar tendon (modelled with shell elements), knee capsule (modelled with shell elements), and flesh (modelled with hexahedral elements). In the model, nodes on the external surface of the bones and nodes defining skin were used to define Delaunay mapping of flesh and other soft tissues (tendon, ligaments, menisci).

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 and are later used for repositioning the knee joint.

Tibiofemoral Motion:

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

The studies on tibiofemoral kinematics have been reported for the in-vivo (Li et al. 2007, Johal et al. 2005, Hill et al. 2000, Asano et al. 2001, Asano et al. 2005) as well as for the cadaveric knee (Elias et al. 1990, Hollister et al. 1993, Churchill et al. 1998, Iwaki et al. 2000, Most et al. 2004, McPherson et al. 2005). These studies have reported relative movement between the tibial and femoral surfaces with respect to different anatomical reference points.

Even though a broad qualitative agreement is observed amongst data reported with regard to type of motion, a quantitative comparison on knee joint kinematics might be difficult between different studies. This is due to the differences in the activities and population studied; the variation in the references

used for measurements and also due to the differences in the way data is reported (rotations or translations). It is indicated that the tibiofemoral movement is a combination of two motions: (1) rotation of femur with respect to tibia (or vice versa) about a flexion-extension axis (F-E axis) (referred hereafter as pure flexion-extension rotation) and (2) sliding / translation of femoral condyles over the tibial plateau.

There is agreement to the effect that the tibiofemoral rotation (pure flexion-extension) has “no fixed” axis i.e. flexion-extension rotation occurs around an instantaneous axis which sweeps a ruled surface (known as an axode, Mow et al. 2000). This eventually makes it difficult to describe or reproduce anatomically correct pure flexion-extension rotation in knee kinematics studies using mathematical models. However, studies have shown that the motion can be described better while considering F-E axis as a fixed axis (Hollister et al. 1993, Churchill et al. 1998, Stiehl and Abbott 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. Based on these studies, a fixed (single) F-E axis has been located and used in the present study.

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

1. The transepicondylar axis (TEA) is defined as the axis connecting the most prominent points on the lateral and medial femoral condyles or the axis connecting the femoral origins of collateral ligaments (Blankevoort et al. 1990, Hollister et al. 1993, Churchill et al. 1998, Miller et al. 2001). The TEA has also been used in Total Knee Arthroplasty (Berger et al. 1993, 1998).
2. The geometric centre axis (GCA) is defined as the axis passing through the medial and lateral centres of the circular profiles of the posterior condyles (Asano et al. 2001, 2005, Eckhoff et al. 2001, Pinskerova et al. 2001). The circularity of posterior femoral condyles and its application to study knee kinematics has also been reported by Kurosawa et al. (1985), Iwaki et al. (2000), Elias et al. (1990), Hollister et al. (1993) and Churchill et al. (1998). 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 and Pinskerova 2003; Hill et al. 2000; Iwaki et al. 2000). The GCA is shown in Fig. 3 (a). Most et al. (2004) have analysed the

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sensitivity of the knee joint kinematics calculation to selection of the 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.

Studies of tibiofemoral kinematics have shown that the sliding of medial femoral condyle 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° flexion (Fig. 3 (b)). This is also observed by Churchill et al. (1998), Asano et al. (2001, 2005), Pinskerova et al. (2001), Wretenberg et al. (2002) and Johal et al. (2005) and it has been suggested that this uneven femoral sliding can be approximated as an external rotation of the femur about a longitudinal rotation axis (L-R axis) fixed in the medial compartment of the tibia.

- From the above discussion about knee kinematics, the following information can be concluded:
1. Tibiofemoral motion can be approximated by two rotations: Flexion about F-E axis followed by Longitudinal rotation about L-R axis.
 2. Flexion-Extension motion can be approximated by a rotation about a single 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 L-R 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 are shown in Fig. 4 (a). Rotations of the femur about these two axes were used to approximate the knee joint motion. Mechanical and anatomical axes (as described in Luo 2004) of the femur and tibia were also located as shown in Fig.

4 (b). 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).

For repositioning, the femur, tibia and fibula were treated as rigid bodies. For producing knee flexion, the femur was subjected to rotation about F-E axis followed by a rotation about L-R axis while keeping the tibia fixed. The relation between these two rotations was adopted from Asano et al. (2001). As the bones were rotated as per available kinematic information, complete control over the bone kinematics was achieved.

Patellofemoral Motion:

Like the tibiofemoral joint, there are many studies on the patellofemoral joint that investigate various aspects of its motion. Some studies focus on analysis of patellofemoral forces (contact areas / pressure) (Singer et al. 1995, Zavatsky et al. 2004), while few others investigate patellofemoral kinematics (Heegard et al. 1995, Asano et al. 2003, Zavatsky et al. 2004, Li et al. 2007). The data of kinematics is presented in terms of translations (Anterior-Posterior, Medial-Lateral, Superior - Inferior) of the patella with respect to either a tibial reference (Li et al. 2007) or a 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 and Weinstein 2001).

In the present study, data from Zavatsky et al. (2004) has been used to generate patellar motion. In their study, patellar flexion, internal – external rotation and medial – lateral tilt were measured against the tibiofemoral flexion. The coordinate system used for the patella is shown in Fig. 5. Internal – external rotation being less than 2% of the flexural rotation, it was neglected.

Flesh Mapping and Delaunay Tetrahedralization:

Various techniques have been developed by graphic designers to deform structures for animations in games (Sheepers et al. 1997, Aubel and Thalmann 2001, Dong and Clapworthy 2002, Blemker and Delp 2005 and Sun et al. 2000). But most of them either deal with models defined with surfaces or with structures defined by ellipsoids. The model used in the present work includes details of bones, flesh and soft tissues. Hence, such methods cannot be used directly. A new technique has been developed to generate controlled deformations of soft tissues of the FE model.

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This new method is based on mapping of nodes of the deformable components using Delaunay tetrahedralization. Delaunay tetrahedralization (Preparatta and Shamos 1988) is a method to partition the space formed by a point cloud data into a set of tetrahedrons. The space between the skin and the bones is thus partitioned into a set of tetrahedrons, and the volume coordinates of nodes of the deformable components are then found in the tetrahedron they lie in. Once the bone is transformed through affine transformations defined above and the skin is transformed through a set of geometric heuristics, these volume coordinates are used to get the new position of the nodes of the deformable components.

Implementation of the Methodology On Lower Extremity (Knee Joint):

A code using VC++ (programming language and GUI) and OpenGL (Graphics platform) has been developed to handle both the processes viz., affine transformations of the bones and mapping of the soft tissues. The steps of the repositioning algorithm are shown in the flow chart in Fig. 1 and have been detailed above.

In the first step, contours of the skin were identified as shown in Fig. 6. A total of 86 skin contours were identified in the initial configuration of the given model. The soft tissues (hexahedral elements) nodes are contained in the space between the skin contours and outer surface of the bones. Delaunay tetrahedralization was then carried out and tetrahedrons were generated using nodes on the outer surface of bones and skin nodes. The process does not alter the connectivity or type of the soft tissue elements. The tetrahedrons were used for the mapping the soft tissue nodes using volume coordinates. This mapping information was later used for reverse mapping to obtain new position of soft tissues. The bones were then given affine transformations (flexion followed by longitudinal rotation as per data described earlier in Asano et al. 2001). The skin contours were also then transformed.

A schematic diagram of the model with skin contours and bones (body 1 region above knee joint and body 2 region below knee joint) is shown in Fig. 7 (a). If the skin contours were given the same affine transformations as the bones, the skin contours would penetrate into each other as shown in Fig. 7 (b). These penetrations were removed by rotating each contour about an axis lying in the plane of contour along the medial-lateral direction. This rotation is termed as *parallelizing rotation* as it orients the contours in the joint vicinity to become almost parallel to each other. One such typical contour in the region above knee joint, along with its axis and direction of rotation is shown in Fig.

8(a). Positions of contours after parallelizing are shown in Fig. 8(b). All the contours are not rotated by the same amount. The amount of rotation reduces to zero as the distance of a contour being rotated increases from the joint. The amount of rotation was calculated as shown in Fig. 10. The link P_1P_0 corresponds to the line passing through the centre of skin contours of the thigh region and link P_2P_0 to the line passing through the centre of skin contours of the calf region. P_0 is the knee joint position.

$\vec{L}_1$ and $\vec{L}_2$ are the vectors along the inferior – superior direction of upper leg and lower leg respectively. Suppose the angle between $\vec{L}_1$ and $\vec{L}_2$ (flexion angle) is 2θ . Let a typical contour in region P_1P_0 be denoted by 'i'. The normal of plane of i^{th} contour is then set as follows, where the orientation of the contour at the joint is set to be the bisection plane of the joint.

The algorithm is based on Jianhua et al. (1994), which describes the technique for surface models used in computer animations. The algorithm shown in Fig. 9 is used in the present study to move the skin nodes to the new position.

Thus the amount of rotation of a plane reduces as its distance from the joint increases and at ends it becomes zero and a kink free repositioned model with none penetrating contours is obtained. But this operation reduces the distance of skin nodes from the bone (link P_1P_0 or P_2P_0 depicting bones in Fig. 10) they are attached to, resulting in a reduced volume of flesh. To prevent this volume squashing (in other words, to ensure volume conservation), the skin contours were scaled out so that distances of points on contours from the attached link remain constant. After transformations of the i^{th} contour (with the centre at Q_i) to the new posture, a point A on this contour (in Fig. 10) is located. After rotation as in Fig. 10, the point A is transformed to A_1 (Fig. 10). In the initial posture, the distance of point A from the link is r_0 (Fig. 10). The point A_1 is scaled to position point A_2 in the direction $\vec{Q_iA_1}$. The scaling factor is calculated from Eq. (1). The Eq. (2) gives coordinates of point A_2 .

$$\text{scale_factor} = \frac{r_0 \parallel \vec{Q_iA_1} \parallel}{\text{dist}(A_1, P_0P_1)}, \text{ Hence} \quad (1)$$

$$A_2 = Q_i + \text{scale_factor} \times (A_1 - Q_i) \quad (1)$$

The effect of the parallelizing and volume squashing operations on the model is shown in Fig. 11. As marked in Fig. 11(a), contours in the vicinity of the joint penetrate into each other after flexion. These penetrations are removed by parallelizing as can be seen in Fig. 11(b). The calf portion

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becomes lean in the repositioned model due to the parallelizing (which reduces its volume). To compensate for this reduction and regain the shape, volume preservation scaling is applied as shown in Fig. 11(c). Once the final position of bones and skin nodes is achieved, the tetrahedrons generated in the initial configuration of model, are relocated using the saved information after tetrahedralization of the initial model. Nodes of the soft tissues are inside the tetrahedrons and are then mapped to the new position using the known volume coordinates in their respective tetrahedrons. The repositioned model is thus obtained. As the 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 the initial model has penetrations). But due to the 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, come in contact after flexion) penetrations were observed. To remove these penetrations the skin contours were locally stretched. This stretching of the skin is also observed when the knee joint is flexed.

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 mesh quality parameters are found to be of poor quality compared to their respective values in initial model.

Mesh Smoothing:

To improve the quality of the mesh at higher flexion angles, a new mesh smoothing algorithm has been developed by modifying standard Laplacian mesh smoothing algorithm (Hansbo 1995). The algorithm is a more generic form of the modified Laplacian algorithm demonstrated by Khattri (2006) for structural meshes. The algorithm can operate on structured as well unstructured 2D and 3D meshes. In order to improve the element quality, the algorithm displaces internal nodes (nodes on the boundary of any components are kept intact) in the space without changing the number of nodes / elements. While moving an internal node, the algorithm ensures improvement in the quality metric parameters of other elements sharing that node. It also preserves element connectivity.

SIMULATIONS WITH REPOSITIONED MODEL

In order to evaluate the numerical stability of the repositioned model, the model was subjected to dynamic finite element simulations without any mesh editing. A steel impactor of 150 mm diameter and weighing 20.1 kg was impacted on different regions of the model at a velocity of 5 m / s in a configuration similar to that used by Yoganandan et al. (2001) (Fig. 14). The impactor surface was modeled to be padded with the Rubatex (R-451) foam of 32 mm thickness defined as viscoelastic material.

In two of the configurations (Fig. 14 (a) and Fig. 14 (b)), impact was in the thigh region from the anterior (frontal) and lateral directions. In two other configurations the knee joint and the region below the knee joint were impacted from the frontal direction (Fig. 14 (c) and Fig. 14 (d)).

In each case, the simulation was run for 100 ms. No anomalies (errors / warnings / termination) were encountered during the simulation. For all the four impact configurations, the maximum hourglass energy was less than 10% of the total energy. The maximum hourglass energy was observed in case of impact on the knee flesh and was found to be 3.05% of the total energy. This was anticipated as the element distortions are maximum in this region. These results suggest that numerical anomalies are not generated in the repositioned model and it is suitable for the dynamic FE simulations in the new posture. The results obtained for these simulations and their comparisons with experimental data are not being presented here as that is beyond the scope of the current paper.

RESULTS

The complete process of repositioning takes approximately 104 seconds to reposition the given FE lower extremity model. The bones (tibia / femur and patella) are repositioned using affine transformations and hence the method allows complete control over the kinematics being followed. Soft tissues like muscles, ligaments, tendon and flesh are mapped on to the repositioned bone using Delaunay triangulation. The model repositioned with the method developed is shown in Fig. 12. At this stage, the flexion up to 90° has been implemented. As it can be seen in Fig. 12, the repositioned model is free from uncontrolled distortion of model components. The mesh quality is hence maintained.

The mesh quality parameters in the initial model (Min. Jacobian: 0.21, Max. Aspect Ratio: 10.76, Max. Warpage: 166.67, Max. Skew: 75.3) were the targeted quality values for the repositioned model. The minimum time step of the initial model was 1.888E-04 sec and this was maintained for the repositioned model throughout the flexion range. The mesh quality parameters were tracked

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throughout the flexion range i.e. from 0° to 90°. For the flexion up to 90°, the minimum Jacobian (0.21) of the repositioned model is maintained at the initial value, while the maximum warpage (158) was better in the repositioned model compared to that of the initial model. A small degradation of the element's aspect ratio (maximum 25) and skew (maximum 84) was observed. But the number of elements with maximum aspect ratio greater than that in the initial model was minimal (less than 0.05%). Up to 60° of flexion, only two elements have maximum aspect ratio greater than 10.76 (as in the initial model) while at 75° there are only 9 elements (out of ≈ 19000 solid elements) whose aspect ratio was more than 10.76.

A detailed look at the repositioned model revealed that at higher flexion angles the elements in the initial model having a poor quality have degraded more than the other elements. This is not surprising as these elements are in the regions having a complex geometry.

As mentioned in the methodology, a mesh smoothing algorithm has also been implemented to improve the mesh quality of repositioned model especially after 75° of flexion. The smoothing improves the mesh quality, even though further improvement is possible at 90° flexion and is being looked into.

The deformation of the soft tissues generated during the morphing process is not random but reflects actual anatomical behavior. This is confirmed by the analyzing the collateral ligaments at various flexion angles which were repositioned to the new flexion angle by mapping along with the other soft tissues. It is observed that during the flexion the anterior fibers of collateral ligaments are stretched while the fibers on the posterior side remain slack. This behavior of collateral ligaments during the repositioning is consistent with the one reported by Park et al. (2006). As shown in Fig. 13 (a), the most anterior fibers of LCL and MCL are most stretched while the most posterior fibers remain slacked during the flexion. Similar behavior of the LCL and MCL in the model can be observed in Fig. 13 (b) and Fig. 13 (c).

DISCUSSION

Unlike, repositioning with FE simulations (Parihar 2004, Jani et al. 2009), where the repositioning takes a considerable amount of time (as reported, few days to reposition the one leg from 90° flexion to 0° flexion), the method developed in the present study can reposition a given FE model in a few seconds (at present 104 seconds for the same model). The element type, element connectivity and mesh quality of the model are preserved during the process in order to maintain its suitability for

simulations in the new posture. Also, the method presented does not require any precalculated FE Model positional data as needed in the tool reported by Vezin and Verriest (2005).

The method operates with geometric operations like transformations and uses anatomical information (for instance F-E and L-R axes are located based on the anatomical information). This makes the method independent of the coordinate systems used to define the geometries or being used in a particular study. Also, the affine transformations of bone permit implementation of the available data of relative bone movements. The method allows control of the soft tissue deformation and prevents distortions as could be the case with model repositioned using FE simulations (Jani et al. 2009). Due to the controlled deformation of soft tissues, the repositioned model does not require remeshing and can directly be used for further applications.

The method has been applied only to the knee joint and its suitability to other joints needs to be evaluated. At present, the method fails to maintain the mesh quality to expected level at higher flexion (at flexion more than 75°) angles. The obvious reasons for this are (1) significant geometrical changes occurring in the model and (2) presence of elements with poor mesh quality in the initial model. Improvements are being made in the mesh smoothing algorithm to overcome this problem by combining it with optimization based smoothing. Also, as the positional accuracy of repositioned bone depends on the axes of rotation chosen, it is essential to locate axes using anatomical landmarks as accurately as possible.

CONCLUSIONS

A methodology to reposition an existing FE-HBM in a given posture (occupant or pedestrian) to a new position has been presented in the present study. In general, the method is generic and can be applied to all the FE-HBMs. The suitability of the method is shown on lower extremity of the GM / UVA 50th percentile male model. Though the applicability of methodology has been demonstrated to the knee joint only, it can be easily extended to other joints like the hip and shoulder joints. The prime objective of this study was to develop a methodology to quickly reposition limbs to achieve an anatomically correct position, while maintaining the mesh quality of the repositioned model. Minimization of subjective interventions and ensuring suitability of the repositioned model for simulations in new posture without re-meshing was also targeted.

The method is capable of including bone and ligament kinematics data. Hence, the limb positions of the repositioned model are more accurate than the one produced with methods which

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cannot directly include such data. The method efficiently controls the shape of deforming soft tissues and prevents bone to bone or soft tissue to bone penetrations. Also the volume preservation technique introduced controls the volume squashing.

The mesh quality of the repositioned model is maintained as close as possible to that of the initial model. The repositioned model does not require any re-meshing and can directly be used in subsequent simulations. In the new posture also, the model was found to be suitable and stable for dynamic FE simulations. To preserve the mesh quality during the repositioning, a mesh smoothing algorithm is developed and applied. However, with the present smoothing algorithm, the mesh quality could not be maintained at the level of the initial model for flexion angles higher than 75°, the mesh smoothing algorithm is currently being improved to take care of the same.

One of the main limitations of the technique is that the method has so far been applied only to the knee joint and its suitability to other joints needs to be evaluated. Being a geometry based approach, ligament / muscle loading (pre-stressing) in the repositioned state (beyond that available from change in lengths) can't be generated. However, if such data is known from literature, it can be incorporated in the algorithm. Also, as the positional accuracy of repositioned bone depends on the axes of rotation chosen, it is essential to accurately locate the axes on the model using known anatomical landmarks.

These limitations are not severely inhibiting and can be easily addressed in future versions of the technique. It is therefore concluded that Delaunay based repositioning offers an effective way of repositioning the FE-HBMs.

References:

Asano T, Masao A, Tanaka K, Tamura J, Nakamura T. (2001) In vivo three dimensional knee kinematics using a biplanar image matching technique. Clin Orthop, 388, 157-166.

Asano T, Masao A, Koike K, Nakamura T. (2003) In vivo three-dimensional patellar tracking on the femur. Clin Orthop, 413, 222-232.

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.

Aubel A, Thalmann, D. (2001) Interactive modelling of the human musculature, In Proc. of Computer Animation.

Berger RA, Rubash HE, Seel MJ, Thompson WH, Crossett LS. (1993) Determining the rotational alignment of the femoral component in Total Knee Arthroplasty using the epicondylar axis, Clin Orthop, Vol. 286, pp. 40-47.

- Berger RA, Crossett, LS, Jacobs, JJ, Rubash HE. (1998) Malrotation causing patellofemoral complications after Total Knee Arthroplasty, Clin Orthop, Vol. 356, pp. 144-153.
- Bidal S, Bekkour T, Kayvantash K. (2006) M-COMFORT: Scaling, positioning and improving a human model, SAE # 2006-01-2337.
- Blankevoort L, Huiskes R, de Lange A. (1990) Helical axes of passive knee joint motions. J. of Biomechanics, Vol. 23(12), pp. 1219-1229.
- Blemker SS, Delp SL. (2005) Three dimensional representation of complex muscle architectures and geometries, Annals of Biomed. Eng., 33, No. 5, 661-673.
- Churchill DL, Incavo SJ, Johnson CC, Beynon BD. (1998) The transepicondylar axis approximates the optimal flexion axis of the knee. Clin Orthop, 356, 111-118.
- Dong F, Clapworthy GJ. (2002) An anatomy based approach to human muscle modelling and deformation, IEEE Visu and Comp Gr, 8, 2, 154-170.
- Eckhoff DG, Bach JM, Spitzer VM, Reinig KD, Bagur MM, Baldini TH, Rubinstein D, 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(4), pp. 96-104.
- Eckhoff DG, Dwyer TF, Bach JM, Spitzer VM, Reinig KD. (2001) Three dimensional morphology and kinematics of the distal part of the femur viewed in virtual reality-Part I, J Bone Joint Su Am. 2001; Vol. 83(2), pp. 43-50.
- Elias SG, Freeman MAR, Gokcay EI. (1990) A correlative study of the geometry and anatomy of the distal femur, Clin Orthop, 260, 98 -103
- Freeman MAR, Pinskerova V. (2003) The movement of the knee studied by magnetic resonance imaging. Clinical Orthopaedics and Related Research, No. 410, pp. 35-43.
- Freeman MAR, Pinskerova V. (2005) The movement of the normal tibio-femoral joint, J. Biomechanics, 38, 197-208.
- Grelsamer RP, Weinstein CH. (2001) Applied biomechanics of the patella, Clin Orthop, 389, pp. 9-14.
- Hansbo P. (1995) Generalized Laplacian smoothing of unstructured grids, Comm. In Num. Methods in Engineering, Vol. 11, 455-464.
- Heegaard J, Leyvraz PF, Curnier A, Rakotomanana L, Huiskes R, (1995) The biomechanics of human patella during passive knee flexion, J. Of Biomechanics, Vol. 28(11), pp. 1265-1279.
- Hill PF, Vedi V, Williams A, Iwaki H, Pinskerova V, Freeman MAR. (2000) Tibiofemoral movement 2: the loaded and unloaded living knee studied by MRI, J Bone Joint Su, pp. 1196 -1198.
- Hollister AM, Jatana S, Singh AK, Sullivan WW, Lupichak H. (1993) The axes of rotation of the knee. Clin Orthop 290, 259-268.
- Iwaki H, Pinskerova V, Freeman MAR. (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 Su 82B, 1189- 1195.
- Jani D, Chawla A, Mukherjee S, Goyal R, Nataraju V. (2009) Repositioning the human body lower extremity FE models, SAE # 2009-01-0922.
- Jianhua S, Magnenat Thalmann N, Thalmann D. (1994) Human Skin Deformation from Cross Sections, Proc. Computer Graphics International '94, Melbourne, Australia.

- Johal P, Williams A, Wragg P, Hunt D, 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. Biomechanics*, 38, 269-276.
- Katchburian MV, Bull AM, Shih YF, Heatley FW, Amis AA. (2003) Measurement of Patellar Tracking: Assessment and Analysis of the Literature, *Clin. Orthop.* 412, pp. 241–259.
- Khatti SK. (2006) A new smoothing algorithm for quadrilateral and hexahedral meshes, *LNCS*, Vol. 3992/2006, Springer.
- Kurosawa H, Walker PS, Abe S, Garg A, Hunter T. (1985) Geometry and motion of the knee for implant and prosthetic design. *Journal of Biomechanics* 18, 487–499.
- Li G, Papannagari R, Nha KW, Defrate LE, Gill TJ, Rubash HE. (2007) The coupled motion of femur and patella during In vivo weightbearing flexion, *J. of Biomech. Eng.*, Vol. 129, pp. 937-943.
- Luo CF. (2004) Reference axes for reconstruction of the knee, *The knee*, 11, 251-257.
- Maeno T, 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*.
- 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.
- McPherson A, Karrholm J, Pinskerova V, Sosna A, Martelli S. (2005) Imaging knee motion using MRI, RSA/CT and 3D digitization, *J. Biomechanics*, 38, 263-268.
- Miller MC, Berger RA, Petrella AJ, Karmas A, Rubash HE. (2001) Optimizing femoral component rotation in total knee arthroplasty. *Clin. Orthop*, 392. 38-45.
- Most E, Axe J, Rubash H, Li G. (2004) Sensitivity of the Knee Joint Kinematics Calculation to Selection of Flexion Axes, *J. Biomechanics*, Vol. 37(11) pp. 1743–1748.
- Mow VC, Flatow EL, Ateshian GA. (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.
- Parihar A. (2004) Validation of human body finite element models (knee joint) under impact conditions, *M. Tech. Thesis*, Indian Institute of Technology Delhi.
- Park SE, DeFrate LE, Suggs JF, Gill TJ, Rubash HE, Li G. (2006) Erratum to "The change in length of the medial and lateral collateral ligaments during in vivo knee flexion", *The Knee*, 13, 77-82.
- Pinskerova V, Iwaki A, Freeman MAR. (2001) The shapes and movements of the Femur and tibia in the unloaded cadaveric 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.*
- Preparata FP, Shamos MI. (1988) "Computational Geometry – An Introduction", Springer Verlag, 2/e.
- Sheepers F, Parent R, Carlson W, May SF. (1997) Anatomy based modelling of the human musculature. *Proceeding of SIGGRAPH '97*, 163-172.
- Singerman R, Berilla J, Davy DT. (1995) Direct in vitro determination of the patellofemoral contact force for normal knees, *J. Biomech. Eng.* Vol. 117, pp. 8–14.

- 1
2
3 Stiehl JB, Abbott BD. (1995) Morphology of the transepicondylar axis and its application in primary
4 and revision Total Knee Arthroplasty, J Arthroplasty, 10, 785-789.
5
6 Sugimoto T, Yamazaki K. (2005) First results from the JAMA human body model project, The 19th
7 International ESV Conference.
8
9 Sun W, Hilton A, Smith R, Illingworth J. (1999) Layered animation of captured data, Animations and
10 Simulations'99, 10th Eurographics workshop proceedings, 145-154.
11
12 Untaroiu C, Darvish K, Crandall J, Deng B, Tai Wang J. (2005) A Finite Element Model of the Lower
13 Limb for Simulating Pedestrian Impacts, Stapp Car Crash Journal; Vol. 49.
14
15 Vezin P, Verriest JP. (2005) Development of a set of numerical human models for safety, The 19th
16 International ESV Conference, #05-0163.
17
18 Wretenberg P, Ramsey DK, Nemeth G. (2002) Tibiofemoral contact points relative to flexion angle
19 measured with MRI. Clinical Biomechanics, Vol. 17, Issue 6, pp 477-485.
20
21 Yoganandan, N., Pintar, F. A., Gennarelli, T. A., Maltese, M. R. and Eppinger, R. H., (2001),
22 Mechanism and Factors involved in Hip Injuries during frontal crashes, Stapp Car Crash Journal,
23 Vol. 45.
24
25 Zavatsky AB, Oppold PT, Price AJ. (2004) Simultaneous In vitro measurement of patellofemoral
26 kinematics and forces, J. Biomech. Eng. Vol. 126, pp. 351-356.
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FIGURE CAPTIONS:

Fig. 1 Methodology Flow Chart

Fig. 2 (a) Lower Extremity Model (b) Details of the knee joint (Lateral View)

Fig. 3 Circles fitted on Posterior Femoral condyles in the sagittal plane and GCA (Lateral View)

Fig. 4 (a) The Flexion Extension Axis and Longitudinal Rotation Axis (Anterior View)
(b) Mechanical and Anatomical Axes of Tibia and Femur (Anterior View), (Jani et al. 2009)

Fig. 5 Coordinate system for patellar motion showing axes for Patellar flexion, External – Internal rotation of patella and Lateral - Medial Tilt, (Adopted from Zavatsky et al., 2004)

Fig. 6 Planes of skin contours

Fig. 7 Skin contours (a) Initial Configuration (b) after flexion

Fig. 8 (a) Parallelizing of contours for a typical contour above knee joint (b) Parallelizing and removed penetrations

Fig. 9 Algorithm of contour parallelizing

Fig. 10 Amount of contour parallelizing and estimation of volume preservation scaling

Fig. 11 Orientation and size of skin contours (a) after flexion (b) after parallelizing and (c) after volume preservation scaling

Fig. 12 (a) Repositioned Model (b) Detailed View of Knee Joint

Fig. 13 (a) Collateral ligament fiber behavior during flexion (adopted from Park et al., 2006)
(b) LCL and (c) MCL behavior in the model during flexion

Fig. 14 Configurations for impact simulation of repositioned leg (a) Frontal Thigh (b) Lateral Thigh (c)
Knee Flesh (d) Below Knee

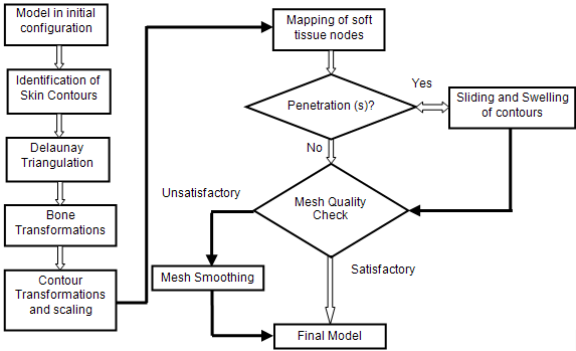


Fig. 1 Methodology Flow Chart

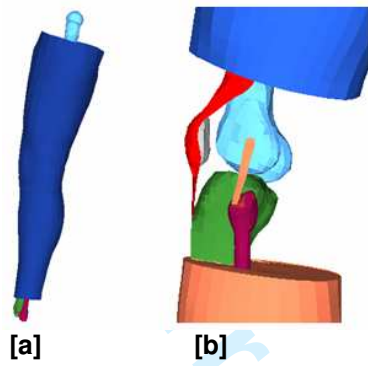


Fig. 2 (a) Lower Extremity Model (b) Details of the knee joint (Lateral View)

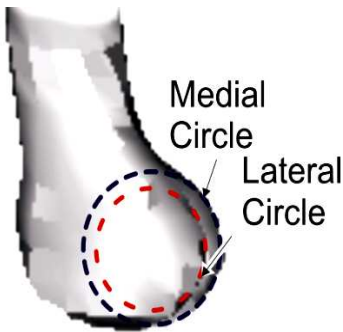


Fig. 3 Circles fitted on Posterior Femoral condyles in the sagittal plane and GCA (Lateral View)

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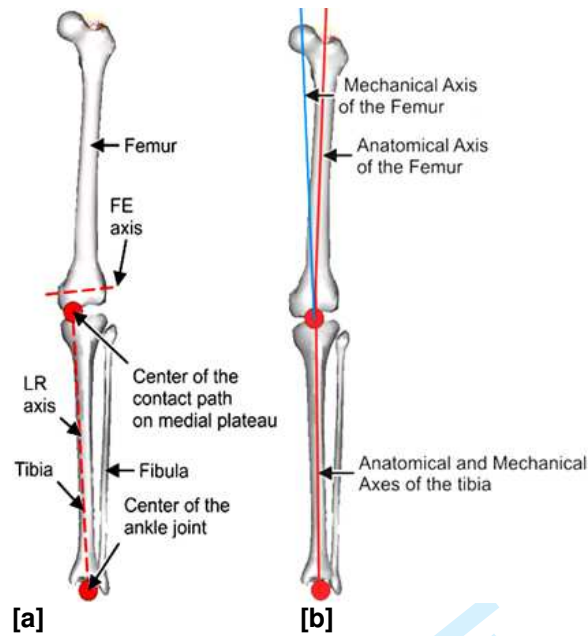


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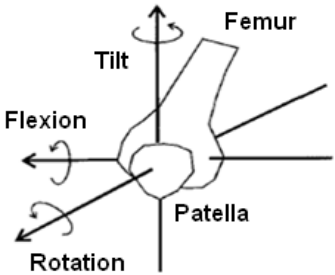


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Traffic Injury Prevention

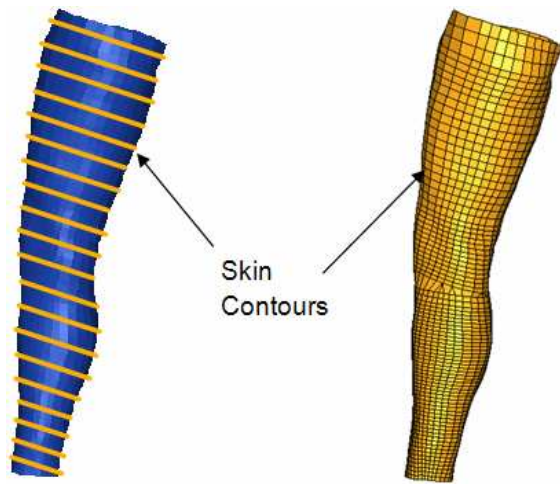


Fig. 6 Planes of skin contours

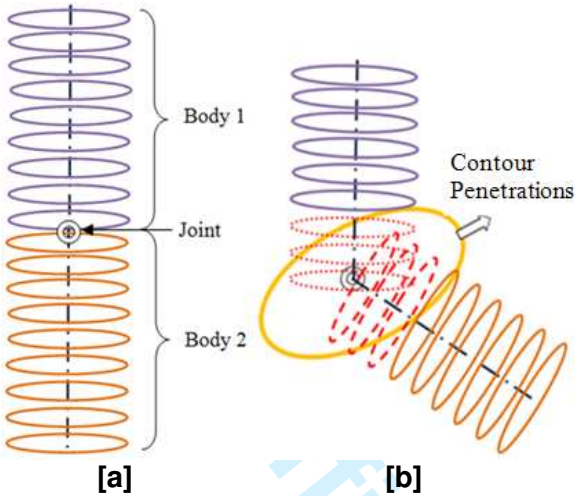


Fig. 7 Skin contours (a) Initial Configuration (b) after flexion

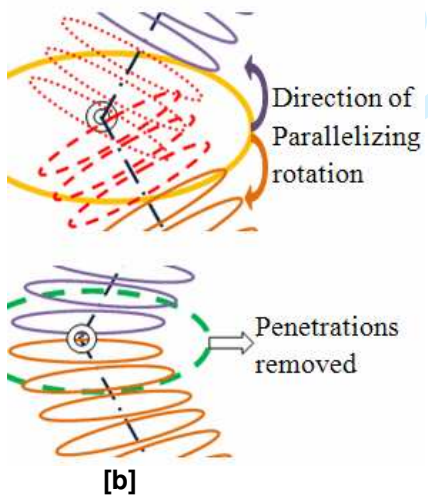
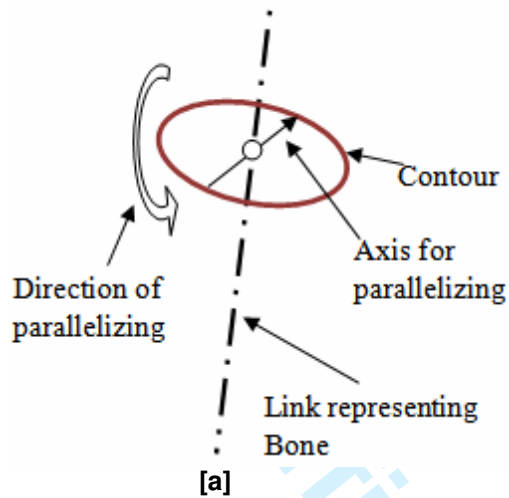


Fig. 8 (a) Parallelizing of contours for a typical contour above knee joint (b) Parallelizing and removed penetrations

For each plane from P_1 to P_0 ,
{
 Obtain centre of rotation Q_i .
 Find the axis of rotation $\vec{L}$ of plane
 as $\vec{L}_1 \cdot \vec{C} \cdot \vec{L}_2$.
 Adjust plane normal N_i by rotating the i^{th} plane
 about axis along $\vec{L}$ with rotation angle given by:

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

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

 where,
 N_{upper} = number of planes on
 P_0P_1 ,
 N_{lower} = number of planes on
 P_0P_2
 *$i = 1$ refers to the plane
 situated between
 proximal end of tibia
 for the lower leg and
 distal end of femur for
 the upper leg. Planes
 on both the sides of
 this plane are given
 maximum rotation
 (flexion angle / 2) to
 orient them to
 become almost
 parallel to this plane.*
}

Fig. 9 Algorithm of contour parallelizing

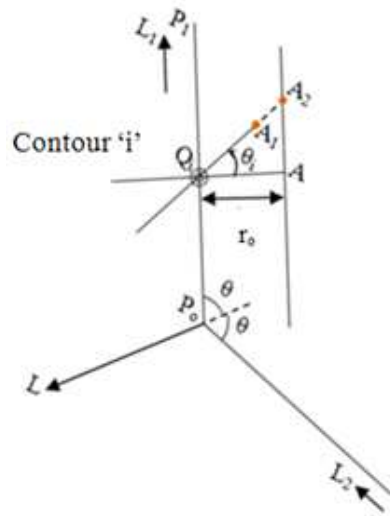
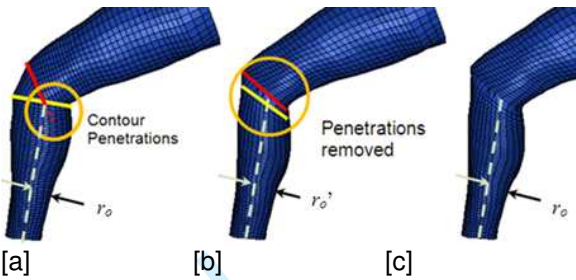
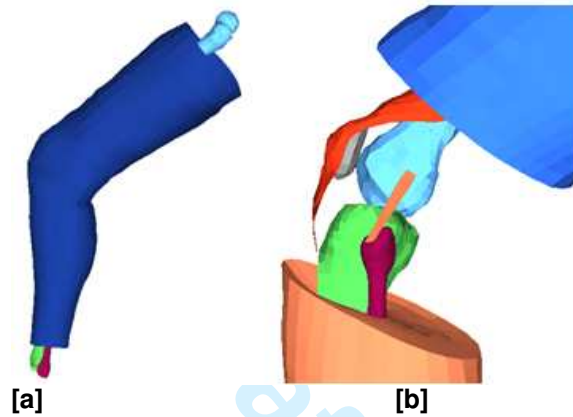


Fig. 10 Amount of contour parallelizing and estimation of volume preservation scaling



[a] [b] [c]
Fig. 11 Orientation and size of skin contours (a) after flexion (b) after parallelizing and (c) after volume preservation scaling



[a] **[b]**
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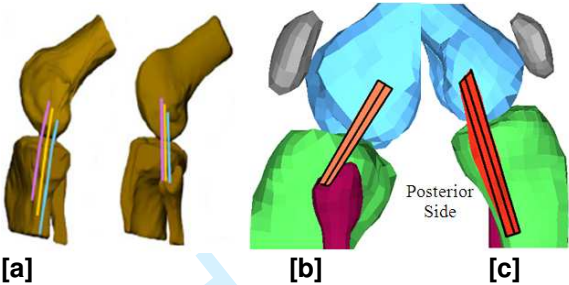


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Traffic Injury Prevention

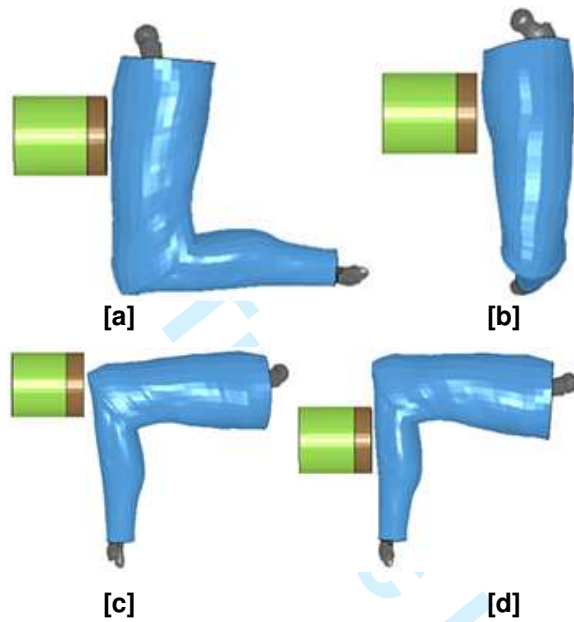


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