Quantification of skin movements artefacts using MRI

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***Abstract*—we present a methodology for anatomical modeling, motion tracking and skin movement quantification of the hip joint using Magnetic Resonance Imaging. First, we capture the subject's anatomy with a static protocol and reconstruct the three-dimensional models of the hip joint. Then reflective markers are attached to the subject's limb and a dynamic protocol is used to track their trajectories. It is known that the skin surface deforms while in motion due to muscle contraction leading to errors in trajectories estimation. In this study, the displacements of the markers relative to the underlying bone are observed and quantified in all three axes. A statistical analysis is performed on the data and a strong correlation is determined between the skin movements and the instantaneous kinematics variables of the joint especially for hip rotations. The quantification of these errors could be used as a basis for a correction procedure for hip joint kinematics.**

Keywords-component; hip joint; MRI, skin aretefacts, statistical analysis;

I. INTRODUCTION

Human joint motion analysis is a prerequisite for various pathological conditions detection and objective evaluation for surgical therapies (cartilage and ligaments deficiencies) as well as non-surgical treatments. Techniques that have been used so far for skeletal movement measurements include invasive methods such as bone pins, external fixation devices and non-invasive methods such as optical motion capture systems. The latter systems use reflective markers attached to the skin and multiple cameras to record the motion of the moving subject. The major limitation of these systems is the skin surface deformation due to muscle contractions.

Estimating and correcting the errors due to soft tissues deformation has been the subject of recent research development. In 1997, Cappello *et al* [1] used an external fixation device and devised a double calibration method to reduce errors associated to non rigid markers' movements. Luchetti *et al* [2] used ad-hoc movements and quantified the skin movement throughout the range of motion. They determined a correlation between the skin movements and flexion-extension and rotation angles. Andriacchi *et al* [3] developed a Point Cluster Technique (PCT), they use a cluster of markers on each segment and an optimal weighting to lessen the effect of skin artifacts. Alexander *et al* expand the PCT and develop the Interval Deformation Technique (IDT) in [4]. They use an optimization method to predict the skin movement for each marker so that to deduce the deformation of the cluster over the recorded time interval.

In this paper, we present a methodology to estimate and visualize bone poses of patient-specific hip joints while minimizing the effects of the skin artifacts using Magnetic resonance Imaging (MRI). More precisely, this research focuses on two aspects: Firstly, we develop a methodology to reconstruct the 3D surfaces of the hip joint as well as its surrounding structures (cartilage and ligaments) from medical imaging: Segmentation techniques are used to interactively generate anatomical 3D surfaces from Magnetic Resonance Images. Secondly, we record the joint motion using a dynamic MRI protocol and reflective markers attached to the subject's limb. The markers trajectories are estimated and the skin movement relative to the underlying bone is quantified using information supplied by both the combination of static and dynamic MRI images.

II. MATERIAL & METHODS

A. Anatomical modeling

At first, tissue-specific MRI protocols need to be used to ensure an optimized acquisition of patient's region of interest. Then, the subject undergoes MRI scanner and the captured images are processed to generate the virtual models of the subject's organs. These steps are detailed below:

1) Image acquisition

Ten healthy adult volunteers (5 females and 5 males) have undergone MRI scanning. The acquisition was performed at the university hospital of Geneva, with 1.5 T Intera station manufactured by Philips Medical systems. Four high-resolution MRI scans of T1-weighted spin echo images and one series of T1-weighted gradient echo images are collected. Repetition time varies from 600 to 3000ms and Echo time is 15ms for bone sequences and 18ms for cartilage. The images have an in-plane resolution of 0.76mm x 0.76mm for the bone sequences and 0.94mm x 0.94mm for the cartilage. The in-between plane resolution varies depending on the anatomical region. The highest resolution is performed in the joint region (2 mm), as this is a crucial region for our study.

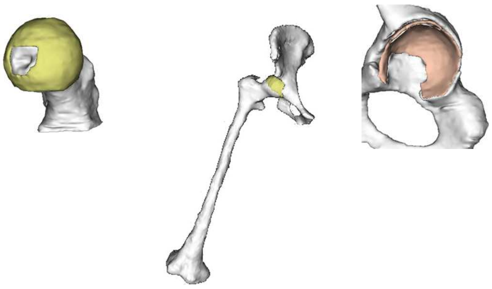

Figure 2. 3D surfaces of femoral head, hip joint and acetabular

2) Anatomical structures delineation

Segmentation is performed using a custom-written discrete snake [5] procedure to extract hip and femoral contours. On each MRI slice, an initial set of points is digitized along each articular curve with a coarse spacing of 1-2cm. The active contour is then used to best fit the actual boundary. This provides an accurate location of the bone contour sufficiently near the initialization curve. In addition to elastic and rigid forces, an image dependent external potential energy, act on each vertex to move it to pixels of interest. Although the snakes have proven to achieve high accuracy while decreasing the time required for manual segmentation, manual corrections are necessary on the slices with fuzzy edges. Prior to the reconstruction process, the segmentation is validated by our medical partner in order to ensure 3D models accuracy.

3) Anatomical structures reconstruction

We use the Visual Toolkit[1] implementation of the Marching Cubes [6] to generate iso-surfaces from the segmented volume. The resultant polygonal surface is simplified with Schroeder decimation algorithm [7]. The decimated polygonal surface is smoothed by adjusting the coordinates of the vertices using Laplacian smoothing “Fig. 1”.

B. Motion Recording

1) Bone and markers pose estimation

As stated earlier, the markers’ trajectories obtained from optical motion capture systems don’t reflect the real trajectory of the bone because of skin deformation. MRI scanners used with dynamic protocols allow the visualization of both markers and bone on the same image making it possible to estimate the markers displacements.

In our study, 10 volunteers have undergone MRI scanner (5 female and five male with a mass mean ± standard deviation of 69 ± 6.05 kg for male and 57 ± 4.07 kg for female and height mean ± standard deviation of 1.72 ± 0.05cm for male and 1.68 ± 0.02cm for female). A static protocol is first used in order to generate the anatomical models of the hip joint of each volunteer. Additionally, a dynamic protocol is used to record the volunteer movements [8]. A cluster of reflective markers injected with contrast agent is attached to the limb of the volunteer who was asked to perform clinical motion patterns: internal rotation, external rotation, flexion, extension, abduction, and adduction. A tracking device is used to ensure the movements repeatability “Fig. 2”.

Figure 1. Horizontal tracking device

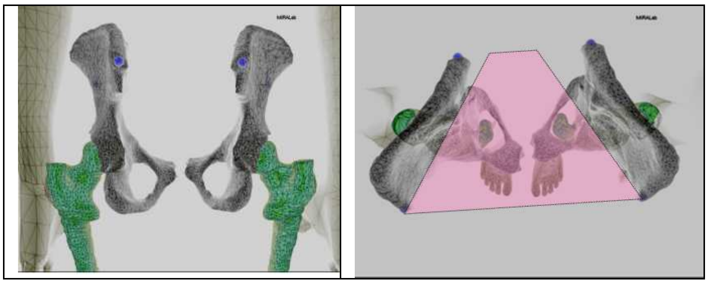

Figure 3. ISB Pelvis Coordinates system: Front View and Top view

The resulting dynamic MRI set provide temporal information on a single slice. On the other hand, the static MRI provides complete 3D volume information of the bone morphology. A 2D/3D registration technique between dynamic and static MRI is developed using functional energy minimization procedure based on similarity metric. More details on the method are given in [9].

Moreover, we use the reconstructed surfaces of the hip joint to calculate the joint center and determine the anatomical axes. The hip joint center is calculated using a functional technique:
The pelvis surface is fixed and the hip joint center is calculated as the point inside the femoral head that remains fixed during joint motions for a restricted range of motions without impingements. More details on the technique could be found in [10]. The hip joint anatomical axes are determined using the calculated hip joint center and the bony landmarks of the 3D surfaces “Fig. 3” according to the recommendations of the Standardization and Terminology Committee (STC) of the International Society of Biomechanics [11].

At this stage, we have for each volunteer in the MRI coordinate system:

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[1] www.vtk.org

- the markers positions for each instant frame
- the hip joint center (HJC)
- the femur embedded anatomical system (FEA)
- the hip embedded anatomical system (HEA)
- The bone and pelvis transformation matrix M_i from the initial position to frame *i*.

We can estimate for each instant frame, the positions and orientation of both the hip anatomical system (HEA) and the femur anatomical system (FEA). Thus, we can compute at each instant frame, the position of the femur relative to the hip. The resulting transformation matrix is converted to Euler angles and linear displacements according to STC of the ISB convention of the hip joint axis

TABLE I. MARKERS DISPLACEMENTS VALUES

	X	Y	Z
Minimum	-14.473	-8.8326	-23.304
Maximum	14.713	3.8837	17.282
Mean	0.847	-1.51	-2.885
Standard deviation	5.33	2.85	6.7
Root Mean Square	5.3502	3.2141	7.2403

Minimum, maximum, mean, standard deviation and RMS values for markers linear displacements in the X-Y-Z directions

The markers displacements relative to the underlying bone are quantified by computing the position and orientation of the markers relative to FEA for each frame as shown in the next section.

C. Quantification of markers displacements

The markers cluster are associated a technical frame as suggested by PCT in [3]. The frame position is determined by the center of the mass of the cluster C (t) while the orientation is defined by the eigen vectors of the inertia tensor I (t) of the cluster at instant frame t.

Let p_i be the coordinates of the marker i at instant t in the MRI coordinate system and n the number of the markers in the cluster, then we have:

$$C(t) = \frac{\sum_{i=1}^{n} p_i}{n}$$

$$I(t) = \begin{vmatrix} \sum_{i=1}^{n} p_{i,y}^2 + p_{i,z}^2 & \sum_{i=1}^{n} -p_{i,x}p_{i,y} & \sum_{i=1}^{n} -p_{i,x}p_{i,z} \\ \sum_{i=1}^{n} -p_{i,x}p_{i,y} & \sum_{i=1}^{n} p_{i,x}^2 + p_{i,z}^2 & \sum_{i=1}^{n} -p_{i,y}p_{i,z} \\ \sum_{i=1}^{n} -p_{i,x}p_{i,z} & \sum_{i=1}^{n} -p_{i,y}p_{i,z} & \sum_{i=1}^{n} p_{i,y}^2 + p_{i,x}^2 \end{vmatrix}$$

Having both the position and orientation of the cluster technical frame and the femur embedded anatomical system FEA) with respect to MRI coordinate system; we can compute the position and orientation of the cluster technical frame with respect to FEA for each instant frame. The cluster technical frame will remain constant if the markers were attached rigidly, but as it is not the case, this will allow us to quantify the markers displacements caused by skin artifacts. The relative movement between the cluster technical frame and the femur is determined using three linear displacements (a, b, c) and three angular (α,β, γ) displacements. Only linear displacements are taken into account in this study. The results of the markers' displacements are reported in table 1 in terms of maximum, minimum, mean ± standard deviation (Table 1).

D. Statistical analysis

For each frame, we have the relative position of the femur and hip expressed in terms of Euler angles and linear displacements.

Additionally, we have the displacements of the markers in the three directions (X, Y, Z). Deriving a precise interdependence between the markers displacements and the movements performed by the volunteers is not possible merely by manual observation. Thus statistical analysis is used. We found that Factor analysis is a powerful tool for achieving this goal [12].

Factor analysis is used in various fields as chemistry, sociology, economics... to estimate maximum likelihood among a set of observed variables. First, a correlation matrix is generated for all the variables then factors are extracted based on the correlation coefficients of the variables. The factors are finally rotated in order to maximize the relation between the factors and the variables.

Following Factor analysis terminology, our data set is made of 6 variables (the three instantaneous kinematics angles of the volunteer movement and the corresponding linear displacements of the markers) and 55 collected observations for each variable. The analysis was performed with 3 factors and oblique rotation.

The factor analysis results are depicted in Table 2 and Fig. 4. It shows that the internal-external rotation movement is highly correlated with the markers displacement in Z-direction while the flexion movement influences the Y-direction.

The next step is the markers displacements modeling knowing their correlation with the movements performed.

E. Skin movements modeling

After testing several interpolators, we found that a quadratic modeling gives the best computational efficiency/norm of the residuals ratio for evaluating the displacements in the Z-direction as a function of internal-external rotation angle whereas the displacements in the Y direction as a function of flexion-extension angle are approximated best with a cubic modeling “Fig. 5 “.

TABLE II. FACTOR ANALYSIS COEFFICIENTS

Variables	Component I	Component II	Component III
α	-0.24722	-0.59719	0.13638
β	-0.43919	-0.16179	-0.19284
γ	0.97134	0.00434	-0.14094
A	-0.05266	-0.11161	1
B	0.093681	1.00	-0.12409
C	0.89956	0.09512	0.09512

We note that γ and C have large coefficients for the component I , they are strongly correlated. The same observation could be made for α, B and component II.

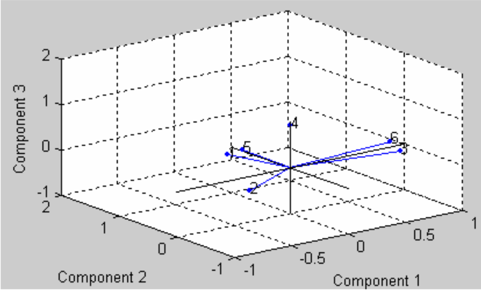
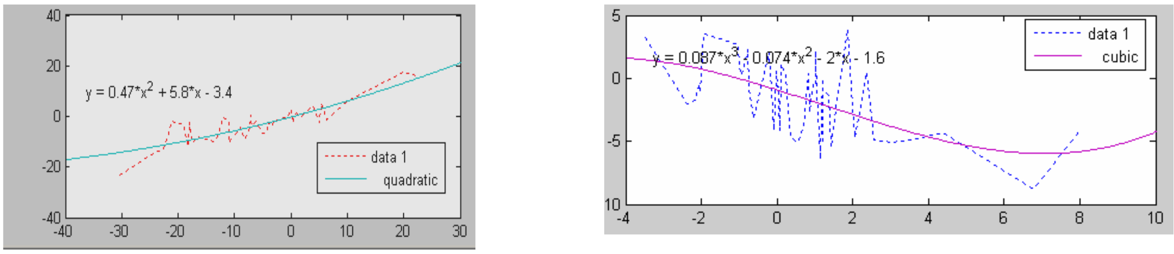


Figure 4. The plot of Factor analysis coefficients. 1-6 correspond respect to α, β, γ, A, B, C



(a) (b)

Figure 5 (a): Quadratic interpolation of the linear markers displacements values in the Z-direction as a function of hip rotation (b): Cubic interpolation of the linear displacements in the Y-direction as a function of hip flexion.

III. DISCUSSION AND CONCLUSION

We have presented a methodology for estimating skin movements' artifacts of the hip joint using MRI dynamic and static protocols. Reflective markers attached to the subject are tracked from the images; their trajectories are analyzed versus bone trajectory and skin movement's errors and quantified. A statistical analysis of the estimated errors showed that the markers displacements in the Z-direction have the most important RMS value. Factor analysis showed the same errors are strongly correlated with hip internal-external rotation while the displacements in the Y-direction are correlated with the hip flexion-extension. The quantification of these errors will be validated and used for a correction procedure.

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