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A 2D 3D registration with low dose radiographic system for in vivo kinematic studies

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***Abstract*—The knowledge of the poses and the positions of the knee bones and prostheses is of a great interest in the orthopedic and biomechanical applications. In this context, we use an ultra low dose bi-planar radiographic system called EOS to acquire two radiographs of the studied bones in each position. In this paper, we develop a new method for 2D 3D registration based on the frequency domain to determine the poses and the positions during quasi static motion analysis for healthy and prosthetic knees. Data of two healthy knees and four knees with unicompartimental prosthesis performing three different poses (full extension, 30° and 60° of flexion) were used in this work. The results we obtained are in concordance with the clinical accuracy and with the accuracy reported in other previous studies.**

***Index Terms*—2D 3D registration, frequency domain, knee joint, unicompartimental prosthesis, low dose radiography.**

I. Introduction

In this paper, we introduce a 2D 3D registration method using 3D data and two 2D radiographs obtained by an ultra low dose radiograph system called EOS. This registration is done in an automated process in the frequency domain using the central slice Fourier transform theorem [1]. Using this technique, the motion of bones or unicompartimental prosthesis can be estimated from different acquisitions during the flexion motion of the knee joint.

In the next section, we present the motion estimation techniques used to evaluate the bones positions and poses. These techniques are generally used for healthy bones as well as for prosthetic ones. In the third section, we present the material we use and our method to determine the positions of the studied object. Finally, in the fourth section, we show the results we obtain evaluating the position of bones and condylar unicompartimental prosthesis during flexion.

II. Motion estimation

To make a motion estimation of the bones, different approaches are used. An approach to acquire these data and to make the motion analysis is to identify some points of interest to determine the pose and position of the whole body. Manual identification of such anatomical points on medical images is generally consuming time. Moreover, it lacks of accuracy and repeatability. Some recent works make this identification automatically [2]. Other techniques implant tantalum markers into the bones and try to find their positions by radiostereometry analysis [3][4]. Such methods provide good accuracy analysis [5]. However, their invasiveness makes a major drawback to their use.

Another approach widely used to search the motion of the bones is based on comparing acquired radiographs of the joint to digitally reconstructed ones (DRR). DRRs are obtained by simulating the imaging machine work and by generating images where 3D data are in different poses and positions. The result corresponds to the pose and position where DRRs match the real radiographs. These methods are similar to ours since they use two radiographs to make a registration with the 3D data.

In such methods, many algorithms were developed to generate the DRRs. First, there are techniques which are based on the ray casting algorithm. This technique simulates the path of the X ray in the fluoroscopy to make a radiograph like images [6], [7]. This generation is a consuming time step and makes a bottle neck in this method of 2D 3D registration. Many authors tried to propose improvements to reduce the calculation time. Birkfellner used projecting voxel method [8] and LaRose reduced the time calculating of DRRs by interpolating some of them from previously calculated ones [9]. Kim and Lacroute used an appropriate hardware to reduce time calculation of DRRs [10], [11]. Others used mathematical transformations to reproduce DRRs, such as cylindrical transform in [12] and Fourier transform in [13]. In a second step, and once the simulation is made, a comparison between the real radiographs and the calculated ones is done. Here also, many measures were used. However, two measure categories can be distinguished: intensity based distances such as correlation, mutual information, and features based distances such as gradient based measure. Penney compared different similarity measures in the purpose of 2D 3D medical registration [14]. He found that the gradient difference and pattern intensity give the best results. Recently, other interesting distances are developed as in [6], [15]. An optimization method is also used in order to reduce the time of search and to converge quickly to the solution. Using this approach, the CT is the most widely used modality to obtain the 3D data. The radiographs are obtained generally by fluoroscopy in oblique [16] or perpendicular [17] directions.

In our work we propose to use a new low dose radiographic system called EOS. This modality provides a pair (frontal and sagittal) of radiographies and permits the 3D surface reconstruction of bones. We develop a 2D 3D registration method well suited to this modality. The originality consists in avoiding the step of DRR generation known as the time consuming step. The method is also generic and can be applied to different joints and shows the feasibility of using EOS in

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pseudo kinematic studies.

III. Material and Methods

EOS modality allows acquisitions in a weight bearing position. Furthermore, it provides us with two simultaneous radiographs of the joint in quasi static acquisitions. A reconstruction of the 3D surface of the bones of the lower limb, based on the Non-Stereo Corresponding Contours algorithm [18], is also available with the software attached to EOS.

In our application, we use the Fourier slice theorem to make the 2D 3D registration [19], [20]. This theorem states that the Fourier Transform (FT) of a linear projection in a direction is equal to the central slice of the Fourier transform of the object in the same direction. Applying this theorem to the frontal and sagittal radiographs, and taking into account the orthogonality between the radiographs, we obtain this system of equations:

$$\begin{array}{rcl} Slice_{\overrightarrow{u}}\left(FT\left(\varphi\right)\right) & = & FT\left(\varphi_{sagit}\right) \\ Slice_{\overrightarrow{v}}\left(FT\left(\varphi\right)\right) & = & FT\left(\varphi_{front}\right) \\ \overrightarrow{u} & \perp & \overrightarrow{v} \end{array} \quad (1)$$

Where $\overrightarrow{u}$ and $\overrightarrow{v}$ are the axes of projections.φ,φ_{sagit},φ_{front} are respectively the 3D object and its sagittal and frontal projections . Hence, we convert the problem here to search directions verifying these equations. Using this theorem, we make our registration in frequency domain. In the first step, only the modulus of the different data is used to make comparison between the Fourier transform (FT) of the radiographs and the slices extracted from the modulus of the FT volume. This leads to searching only the rotation (poses) in a first step. Depending on a previous study [19], we use a global similarity measure d taking into consideration the differences in the frontal and sagittal sides at once. It is defined as:

$$d = \left(\sum_{i,j} \left(I^f_{i,j} - F_{i,j} \right)^2 + \sum_{i,j} \left(I^s_{i,j} - S_{i,j} \right)^2 \right)^{1/2} \quad (2)$$

where F (resp S) is the modulus of the frontal (sagittal) radiograph Fourier transform and I^f (resp I^s) is the modulus of the frontal (sagittal) slice extracted from the Fourier transform of the 3D data. i, j correspond to the frequency bins. We apply an adapted gradient descent method as an optimization method to look for the minimum measure between the central slices extracted from the 3D data and the radiographs. The minimum corresponds to the rotation between the radiographs and the data volume.

Once the rotation found, we search for the translation by using the phase information as a cross spectra between frontal (sagittal) slice extracted as a solution of the rotation and the FT of the frontal (sagittal) radiograph denoted φ_{2D}.

$$Slice_{\overrightarrow{u}} = e^{i2\pi(T_{x1}f_{x1}+T_{x2}f_{x2})} FT\left(\varphi_{2D}\right)$$

Finally, we can easily get the translation components by using: $TF(e^{i2\pi(T_{x1}f_x+T_{x2}f_z)}) = \delta_{T_{x1},T_{x2}}$. And hence, the search of the 3D translation is made in two 2D images in a non iterative way.

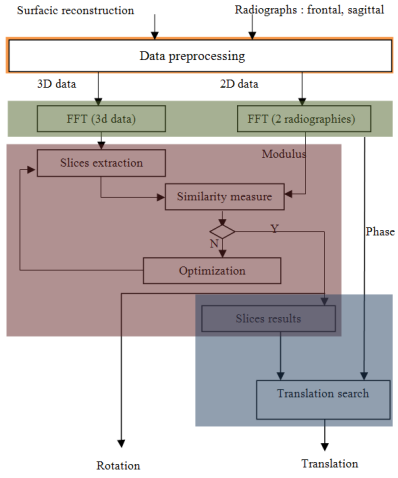


Figure 1: Proposed algorithm scheme

The global algorithm is described in the figure 1. The red part of the scheme represents the search of the rotation and the blue part shows the search of translation.

Preprocessing techniques are added as a first step into the algorithm described above. In this step, we resample the 3D surface envelop data into an isotropic grid corresponding to the sampling of the 2D data. The volume is filled homogeneously to obtain a bone with a unique density. This is different from the real bone density but the results will show that the method is robust to this bias. In this work, we use this method on two different data: prosthesis and healthy bones. Unlike the data of the prosthesis, the 2D data of the long bones (femur and tibia) do not include the entire projections. In fact, we use cropping to focus on the knee joint and select the distal femur and the proximal tibia in the radiographs. This allows us to reduce the time calculating and the memory consuming (2D used data dimensions are 256x256). However, in the frequency domain, this type of cropping creates well known artifacts called leakage phenomenon. Hence, this phenomenon can induce an offset in the rotation results. Moreover, in the 2D data, other difficulties have to be managed such as the presence of noise in the images. Soft tissues and some parts of non studied bones such as patella may perturb results.

To overcome these problems in the healthy bones data, we use a windowing technique to focus on the studied bone. We try three different windows to the bones data in the preprocessing step (figure 1) which eliminate the border effects in the frequency domain caused by the crop of the bones radiographs. Moreover, this window reduces the influence of the soft tissues and the other bones in the images. We make a comparison between the results obtained by the three different windows. The windows used are (1) Hanning, (2) Gaussian and (3) Blackman-Nutall window and can be expressed in one dimension as in [21]:

(1) $= \sum_{i=0}^{1} (-1)^i a_i cos(\frac{2\pi i}{N} n)$, $a_0 = 0.5$, $a_1 = 0.5$

(2) $= exp(-\frac{(n/N)^2}{2\sigma^2})$

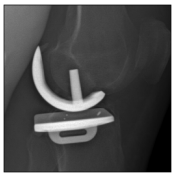
(3) $= \sum_{i=0}^{2} (-1)^i a_i cos(\frac{2\pi i}{N} n)$, $a_0 = 0.42$, $a_1 = 0.5$, $a_2 = 0.08$

Since the dimensions of the prosthesis are smaller, the

windowing step is not useful and will not be applied. In fact, the projections of the prosthesis are contained into the images. To eliminate the effect of the presence of bones and soft tissues in the radiographs, we use the high intensity of the prosthesis to make a threshold of it. This high intensity is generally observed with prosthesis. It is caused by the metallic nature of their manufacturing materials. The results are presented in the next section.

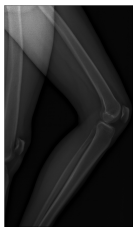
IV. Results and discussion

We apply the method described above on radiographs of patients with unicompartimental prosthesis of the knee joint, and on radiographs with healthy knees. Each patient makes three pairs of acquisitions : in full extension 0°, 30° and 60° of flexion. For data with prosthesis, we use the model provided by the Oxford Company as 3D data in the initial position. For healthy knees, we use the reconstructions of the initial position (knee in extension). In figure 2, we show a sagittal radiograph of a prosthesis (a) and its 3D surface (b). In figure 2(c), we show a lateral radiograph of a lower limb without prosthesis, and we show the reconstruction of the femur in (d).

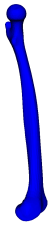


(a) EOS sagittal radiography of the prosthesis

(b) 3D reconstruction of the prosthesis



(c) EOS radiograph of lower limb



(d) 3D reconstruction of a femur

Figure 2: Entry data used for registration

Initial images are with a resolution of 0.18x0.18 mm^2. To reduce the memory cost and the run time, we modify this resolution by interpolating the sagittal and frontal data to obtain images with new resolutions. The results of the rotations R (R_x, R_y, R_z) and the translations T (T_x, T_y, T_z) are respectively expressed in degrees and mm in a global 3D coordinate system $(\vec{x}, \vec{y}, \vec{z})$.

To validate the method, we compare the results obtained by our algorithm to those obtained by a rigid manual registration using the software associated with EOS to generate simulated radiographs. This software takes into account the calibrated space of the acquisition to determine the 3D position of different points manually identified, allowing the knowledge of the 3D position of the object (prosthesis or bone).

In vivo healthy joint:

Data of two healthy subjects are with a new resolution of 1x1 mm^2. In tables Ic and Ib, we show the translations and rotations results obtained using different windows. For these tests (figure 2c), the positions of the in vivo bones are searched from the initial reconstruction in the extension position. The motion reference between 0-30° and 0-60° of both subjects (F1,T1) and (F2,T2) is shown in tables Ia and Ic. In tables Ia, using these in vivo bones, the rotations vary from -42.8° to 34.2° at the different components. Comparing the different windows in Ib, the (2) windowing technique shows the best results where the rotation errors are less than 2.7°. For the (1) window and (3) window, these errors could reach 4.3° in some rotation components. For the (2) window, the mean absolute error is 1.45°. Concerning translations, the components vary from -78 mm to 59 mm. For the search of the translational component to be successful, it is required to first find the rotational component with good accuracy. For this, only the second window results are retained to search the translation as it is shown in Ic. Comparing to the reference translations, the mean absolute error is 1.5 mm and the maximum error is of 2.5 mm.

	Software		
	Rx	Ry	Rz
F1	0,8	-20,8	16,5
	-2,4	-38	17,7
T1	3,3	22,8	6,2
	0,4	34,2	1,6
F2	0,2	-30,3	15,8
	-0,8	-42,8	16,5
T2	-2,2	18,7	-3,3
	-5,7	24,1	-16,5

(a) Bone rotation reference

	window1			window2			window3		
	Rx	Ry	Rz	Rx	Ry	Rz	Rx	Ry	Rz
F1	1.3	-20.5	18.2	-0,8	-23	16,5	-3.4	-19.4	19.4
	-3.4	-37.1	17.8	-2,9	-40,4	17,4	-4.5	-37.1	17.1
T1	1.7	25	4.5	6	20,1	6,6	3.4	25	3.4
	1.8	35.4	4.6	2,4	33	2,5	2.3	32.7	4.4
F2	4.5	-33.8	11.4	-0,4	-28,6	18,6	-4.5	-32.7	17.1
	0.57	-42.5	13.7	0,9	-44,6	18,5	0.1	-42.5	14.8
T2	-5.7	22.7	-5.7	-0,8	18,4	-2,9	-4.5	19.3	0
	-5.7	23.8	-12.5	-3,4	24,9	-14,3	-3.4	23.8	-20.5

(b) Results of bone rotations with different windowing

	Software			window2		
	Tx	Ty	Tz	Tx	Ty	Tz
F1	-20	-30	-4	-20	-32	-6
	23	-10	-29	21	-9	-27
T1	-48	-13	36	-46	-11	38
	-3	-15	-9.5	-2	-14	-12
F2	-79	-30	-7	-78	-30	-9
	-51	-34	7.5	-49	-32	10
T2	-49	60	29	-50	59	31
	-28	12	-3	-28	14	-5

(c) Reference and results of bone translations obtained with window2

Table I: Results with bones data

Prosthetic joint:

Data is with a resolution of 0.36x0.36 mm^2. In these tests, the windowing technique is not applied as explained above. Using the prosthesis data (figure 2a), we present the results

we obtained with prosthetic knees and we also present the reference motion in table II: the rotation components of the prosthesis of four patients vary from -65.6° to 9° and the translation components vary from -30.6 mm to 34.9 mm. The maximum error is -1.7° for rotations and -1.44 mm for translations.

Software			our algorithm		
Rx	Ry	Rz	Rx	Ry	Rz
-0.6	-9.4	-5.6	-0.5	-9.1	-4
-1.0	-45.2	-2.2	-2.1	-45.9	-1.5
-0.8	-64.3	5.7	-2.1	-65.3	6.8
0.9	-7.3	0.3	0.5	-7.3	1.1
4.3	-49	0.7	4.5	-49.6	2
1.6	-65.6	2.3	2.1	-66.4	2.2
-3.4	-42.7	9	-4.4	-43.5	10.3
-5	-56.3	7.4	-5.4	-56.6	6.8
-4.7	-35.9	-5.6	-3	-34.7	-4
-7	-53.7	-10	-5.4	-54.9	-9.8

Software			our algorithm		
Tx	Ty	Tz	Tx	Ty	Tz
8.6	8.2	3.9	8.2	8.6	3.6
-15.8	-4.3	-5.4	-15.4	-4.3	-5.4
7.9	9	19.4	7.5	9.7	19.8
-3.6	-8.6	-11.5	-3.6	-9.3	-11.5
31.3	5.4	10.4	31.6	5.4	10.8
6.4	20.8	14.4	7.9	21.2	14.7
-0.7	-9	-28	-0.3	-8.2	-27.7
7.9	-18.7	-14.4	7.2	-19.4	-15.1
1	-1.8	34.9	0.7	-1.8	35.2
17.2	6.4	-30.6	16.9	5.7	-29.8

Table II: Rotations and translations of prosthesis data

In general, the accuracy we obtain remains in the threshold fixed in other previous works. In [22], authors used generation of DRR technique to make a 2D 3D registration for knee bones and reported errors of 2° and 2mm. With these tests, we get good results using our method with prosthesis. Using bone data, we have to use a windowing technique. Comparing three different windows, the Gaussian one provides the best results.

V. Conclusion

In this work, we presented a new method for 2D 3D rigid registration for motion estimation in orthopedic applications. Our method uses a new modality (EOS) allowing low dose radiation for data acquisitions. It is based on the search of a rotation and a translation in frequency domain between a pair of radiographs and a 3D reconstruction in an initial position. We use this method with the femur, the tibia and prosthesis. This shows that it can be used without a major variation with different data and different joints. The errors obtained when using our method are acceptable in such applications. They are about 2° in rotation components and about 2 mm in translation components. In future works, we will apply this generic method to more data from patients and healthy subjects for ligament laxity studies. We will also use it to other joints such as the hip and the elbow.

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